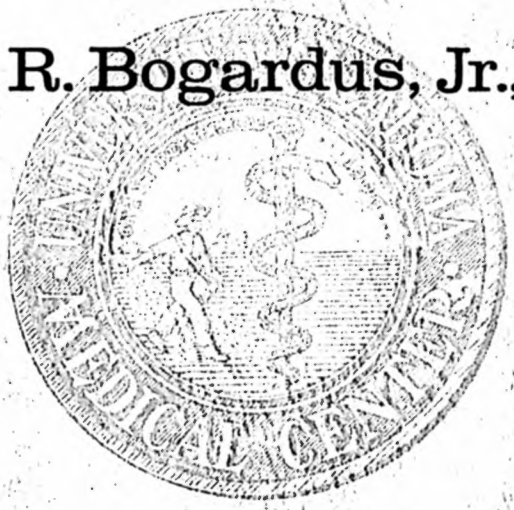
Virgil E. Yoder

PRACTICAL APPLICATIONS of PHYSICS to RADIATION THERAPY

Carl R. Bogardus, Jr., M.D.



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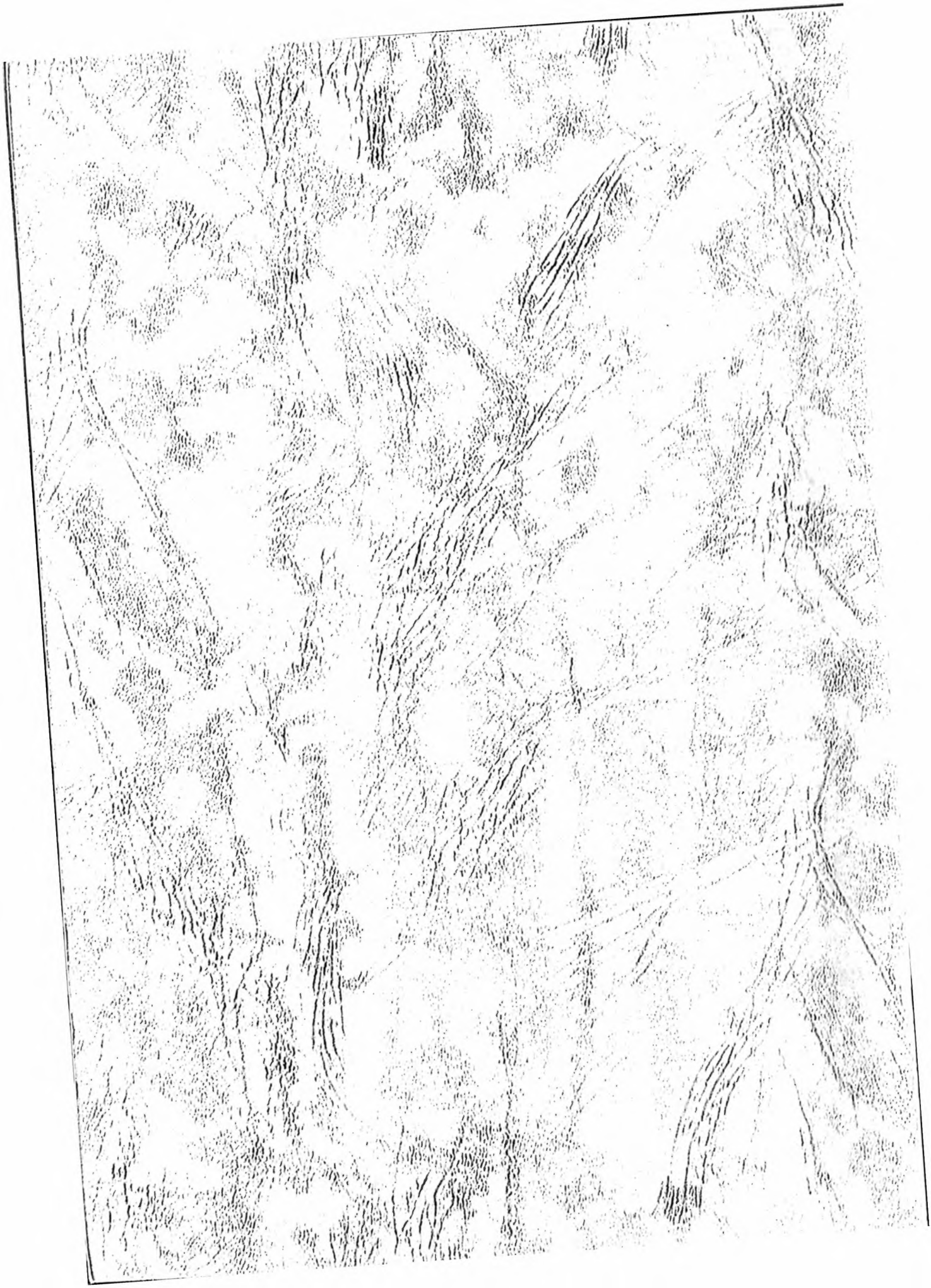

PRACTICAL APPLICATIONS OF PHYSICS TO RADIATION THERAPY
2nd EDITION

Presented as a refresher course
Radiological Society of North America
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In this book, as in the first edition, I have not given a highly detailed discussion of the various phenomena that are found in Radiation Physics. I have limited my discussion to the every day problems that arise regarding physics as encountered by the radiation therapist. I have explained briefly these various phenomena, their methods of action, theory of operation, as well as their practical applications to the every day work. A number of the illustrations and examples have been condensed from the books found in the bibliography. For more complete technical descriptions the major reference books of the bibliography should be consulted.

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I ATOMIC STRUCTURE AND RADIATIONS

A. Properties of Elementary Particles and Atomic Structure.

1. The Atom

Definition: The atom is the smallest unit of an element which retains all of the chemical properties of the element. There are 91 elements existing in nature at the present time. Technetium 99, due to its short half life, does not normally exist in nature. Twelve elements have been man made. The atomic weight or the atomic mass is based on a weight of 16.0000 for oxygen 16. It should be noted that this is the normally accepted mass of oxygen as based on the chemical scale. A second value scale, the physics scale, utilizes the oxygen 16 isotope of oxygen, realizing that oxygen 17 and oxygen 18 do exist. The mass of oxygen 16 is based on the physics scale as equal to 16.000,000,00. The atomic masses of isotopes are based on this physics scale. Hydrogen is the lightest element with a mass of 1.0080, and uranium 238 is the heaviest natural element. Element No. 103, Lawrencium, is the heaviest man made element with a mass no. of 267.

2. The Nucleus

Definition: The nucleus is the central core of an atom composed of neutrons (charge equals 0) and protons (charge equals +1). The preponderance of the weight of the atom resides within the nucleus. The various nuclear particles loose their specific identities when they are combined within the nucleus. They exist only in specific arrangements which are characterized by definite, discrete energy levels.

3. The Proton

The proton has a mass of 1.007597 (physical scale) atomic mass units. All protons are identical; however, each element has a different number of protons. The number of protons found within the nucleus of any specific element determines the atomic number (Z) of the element ${}_{Z}X^{A}$.

4. The Neutron

The neutron has a mass of 1.008986 (physical scale) atomic mass units. The mass of an atom is determined by the total number of neutrons and protons. This is generally designated with the symbol "A" (${}_{Z}X^{A}$). Isotopes are varieties of an element, which have the same number of protons but which have a different number of neutrons; hence, same Z, different A. All isotopes of an element have the same charge and chemical reaction properties.

5. Mesons, Pimesons and Nuclear Binding Energy

The atomic nucleus, as we have previously mentioned, is composed of neutrons and protons which apparently exist in a very close relationship. If one considers the electrostatics involved with the concentration of a collection of positively charged particles, it would seem that the nucleus would not be able to hold together.

There are forces that act within the nucleus which tend to hold the entire structure very solidly together. These forces are referred to as close range binding forces or nuclear binding energy. Precisely what this energy force is is not well known even at the present time. It is felt that there is probably a continuous and extremely fast interchange of sub-nuclear particles between the protons and neutrons. This rapid exchange of particles, probably pi-mesons, supply the binding force, also referred to as the exchange force, which tends to hold the entire nuclear structure together. Mesons have often been referred to as "nuclear glue".

In many of the isotopic nuclear disintigrations a proton apparently converts to a neutron plus a positron, and many times a neutron appears to convert to a proton and an electron. It is felt that these reactions probably take place by an exchange through one of the various meson states, often with a loss of a great deal of energy from the nucleus. It is felt that the meson stage plays an extremely important role in nuclear particle conversions.

6. The Electron

Electrons are negatively charged particles which travel in relatively specific orbits around the nucleus much as the planets of our solar system travel in specific orbits about our sun. The number of electrons circulating in orbits around the nucleus of an atom are always equal to the number of protons found within the nucleus when the atom is in a neutral or uncharged state. These electrons originally were postulated in 1913 by Bohr to travel in specific orbits. Bohr had very adequately discribed the electron energy levels or shells and other important factors of the physics of the orbiting electrons.

A quantum mechanical discription of the atom in current usage changes the appearance of Bohr's original model substantially, but does not noticably alter the energy levels or other observable quantities which Bohr originally postulated.

The centrifugal force of the orbiting electrons is balanced by the electrostatic attraction of the nucleus for that electron. The electron, as a rule, remains in the stable allowed path and does not absorb or radiate energy. As a rule the electron shells are able to hold only a certain number of electrons. The filling of the electron shells takes place with a very specific regularity, a pattern to which is easily identifiable, if you examine the periodic chart of the elements.

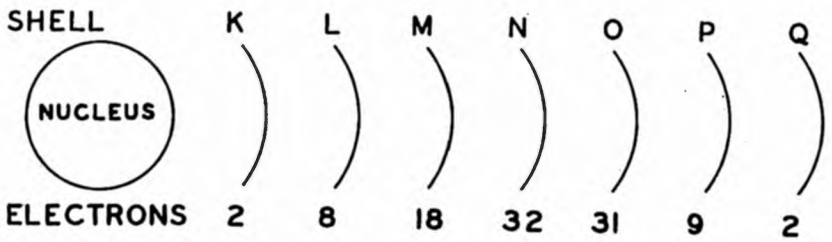


Fig. 1 The Average number of electrons in the atomic shells of Nobelium (No^{102}) The K,L,M, and N shells of most atoms will be relatively constant in the ratios of filling. The O, P, and Q shells fill with a very specific pattern depending upon the classification of the elements.

The chemical properties of an atom are governed by the number of free or valence electrons that can enter into a combination or sharing of electrons with another atom. An ion is an atom that has either more or less electrons than the number of protons found in the nucleus, hence, it is charged, positive or negative.

7. The Positron

A positron is a positively charged electron that has weight identical to an electron. It has a charge that is exactly the same magnitude but of opposite sign. The positron is unable to exist in a free state for any extended period of time. The average half-life of a positron is 10-6 seconds. A free position almost immediately combines with an electron forming a specific compound known as positronium, which then disintegrates by the process of annihilation. (See I-B-6).

8. The Neutrino

The neutrino is a small uncharged particle which, until a few years ago, was only postulated and in fact was thought by few people probably not to exist at all. Its existence, however, has now been proven. A neutrino has the smallest cross-section area of any elemental particle thus far observed or postulated. It should be noted that cross-sectional area is an effective, but not an actual physical area and is used in expressions of interaction probabilities. Neutrinos probably have no identifiable mass. It has been postulated by some that the mass of a neutrino is approximately 1/2000 the mass of an electron, and probably represents a mass of motion or energy. Neutrinos are emitted in beta radioactive decay when electrons are created by neutron conversion into a proton with the creation of an electron and a small bundle of energy, the neutrino. (See section I-B 3,4,5).

Name	Symbol	Mass Unit	Charge	Half-life
Proton	P, H^1	1.007597	+1	Stable
Neutron	N, N^0	1.008986	0	13.2 Minutes average
Mesons	u^+ u^-	.113643(207 x e)	+-	2.15×10^{-6} seconds
	$\pi^+ \pi^- \pi^0$	.149877(273 x e)	+-o	2.5×10^{-8} seconds
Electron	e-	.000548	-1	Stable
Positron	e+	.000548	+1	10^{-6} seconds average
Neutrino	ν	$\leq$ .000001098(1/2000e)	0	Stable
Photon	γ	0	0	Stable
Beta	β	.000548	-1	Stable
Alpha	α	4.0029	+2	Stable

Fig. 2 Properties of the Elemental Particles and Radiations, Mass Units Based On the Physical Scale for Oxygen-16.

B. Radioactive Radiations

Four principle radioactive radiations are of significance at the present time. These are gamma radiation, alpha radiation, beta (positive and negative) radiation, and neutrino radiation.

1. Gamma Ray Decay

Gamma rays are electro-magnetic radiations identical to x-rays except that they originate from within the nucleus, whereas x-rays originate generally from the region of the orbital electrons. Gamma ray emission does not change the nucleus in respect to mass number or atomic number. The difference is that a lower energy level exists within the nucleus following gamma ray emission. The original atom and the residual atom differ in their energy states and are called isomers if the excited state has an identifiable half-life.

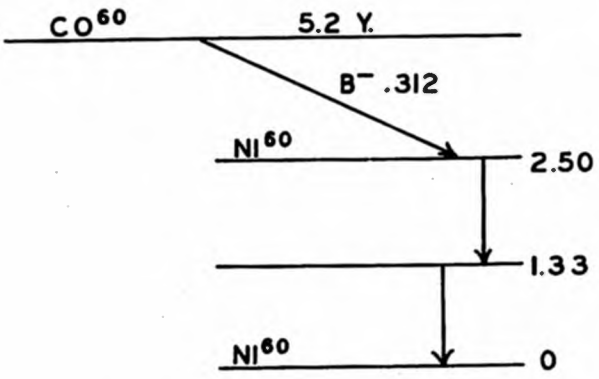


Fig. 3 Simplified Cobalt 60 Decay Scheme - adapted from Radiological Health Handbook (9)

Figure 3 is a simplified version of the disintegration energy and decay scheme diagram for the radioactive decay of Cobalt 60 to excited Nickel 60 to base Nickel 60. In almost 100% of the cases Cobalt 60 decays by beta emission of .312 MeV to Nickel 60 at an excited energy level of 2.50 MeV. From this point it decays successively in two stages with the emission of two gammas of 1.17 and 1.33 MeV ending at base Nickel 60. It should be mentioned that there are two alternative decay routes (not shown) whereby the Cobalt 60 in approximately 0.1% of the cases may decay to the 1.33 MeV level with a beta of 1.48 MeV, and in .044% of the cases decay to a secondary level of 2.15 MeV. Both eventually decay to base Nickel 60.

It is felt that the neutron and the proton exist in a certain configuration within the nucleus and possess a certain inherent amount of energy by virtue of position, much in the same way as the orbital electrons seem to exist in an energy level configuration. When neutrons or protons within the nucleus shift to a different configuration either by virtue of change in sign or dropping to a lower or raising to a higher energy level, energy must be either released or absorbed. The most common method of releasing energy through a transition of this nature is by either particulate or gamma ray emission. This may be best shown by means of a simplified space-energy diagram of the nucleus. (Fig 4)

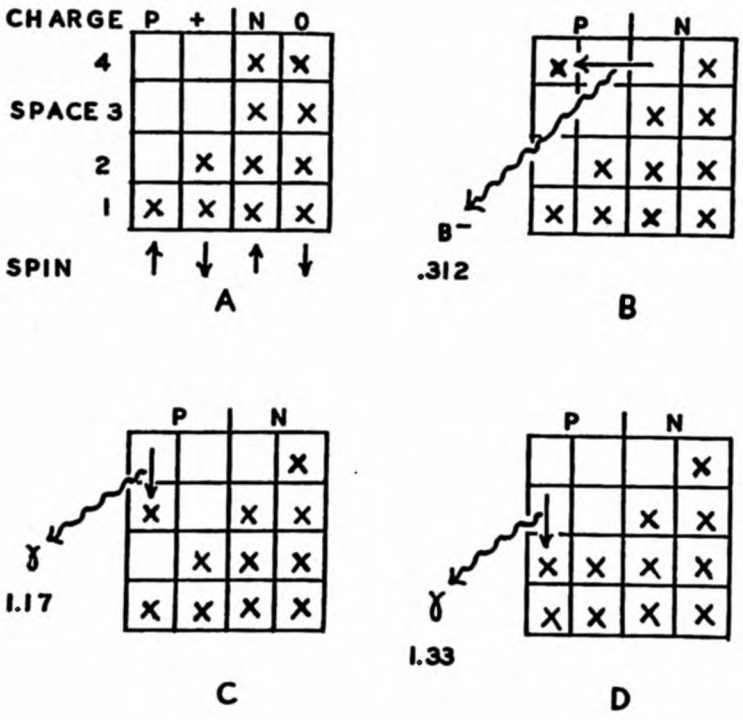


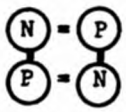
Fig. 4 Space-energy Diagram of the Nucleus

The charge, space area and spin (angular momentum) are different for each nucleon within the nucleus. The "X's" in the diagram (Fig. 4) indicate the various nucleons as they diagrammatically would reside within their particular positions within the nucleus. The neutrons and protons in a nucleus exist in a continuous state of transition through intermediate states by interchange of pi-mesons having positive, zero and negative charge. (I-A-5) Although the basic concept of the conversion of a neutron to a proton is more easily understood and is the overall net result, the actual transition is far more complex and beyond the scope of this discussion. In the case of Cobalt 60. (A), as we have noted before, it decays to excited Nickel 60 by emission of a beta particle as a neutron appears to convert to a proton within the nucleus (B). This transition within the nucleus actually transmutes Cobalt 60 into Nickel 60 by causing a change in the atomic number, note the mass number remains the same. This charge shift releases a small amount of energy in the form of a beta of maximum energy .312 MeV.

The excited Nickel 60 atom then drops the proton to a lower stable energy level by the cascade gamma release on the first level drop of 1.17 MeV (C), and on the second drop of 1.33 MeV of energy (D). This then results in a stable situation within the nucleus.

2. Alpha Particle Decay

An alpha particle is made up of two protons and two neutrons (Fig. 5). This is the same as the nucleus of a normal helium atom minus the surrounding electrons.



$$_{84}Po^{218} \longrightarrow {}_{82}Pb^{214} + {}_{2}\alpha^{4}$$

Fig. 5 Alpha Particle and Alpha Decay Scheme

Alpha emission leaves the nucleus with four less particles. As a rule, alpha emission is one form of decay of extremely high weight isotopes. This is a very important portion of the decay scheme of radium.

3. Beta Radiation or Beta Decay

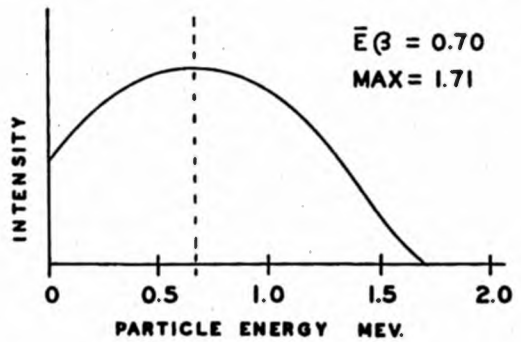
The beta particle is an electron. There are no electrons to be found within the nucleus; therefore, a beta particle or electron is actually created within the nucleus itself. The reaction that takes place is an intranuclear reaction with the net result that a neutron appears to convert to a proton plus a beta particle (Fig. 6) The electron then suddenly finds itself in an extremely hostile environment and is forcibly ejected from the nucleus with varying degrees of energy. A Neutrino is also ejected at the same time and the sum of energies of the electron and neutrino are a fixed value for a given isotope decay. (See section I-B-4), (Fig. 7).

$$N \longrightarrow P^{+} + \beta^{-} + \gamma \qquad {}_{82}Pb^{210} \longrightarrow {}_{83}Bi^{210} + \beta^{-} + \gamma$$

Fig. 6 Beta Decay Scheme

4. Neutrino Production

It was noted that in beta decay a small amount of energy is lost that could not be accounted for by the emitted particle. It was also noted that the beta particles came off with a varying degree of energy or in the form of a spectrum.



It was thought that this energy undoubtedly was lost by means of another particle and the neutrino was thus postulated. It has now been proven that the neutrino does exist. (I-A-8). The neutrino is of no significance to us in radiation therapy at this time.

Fig. 7 Beta particle energy spectrum for Phosphrous - 32.

5. Positron Decay or Production

It was found that some isotopes emitted positive electrons. These are called positrons. The positron is a particle that has exactly the same dimensions as an electron. (I-A-7) It was noted that in these cases a proton apparently converts, by an intranuclear reaction, with the net result that a neutron plus a positron is created. There are no positrons created in the radium decay scheme, however, positron decay is quite important in the field of radioisotopes and should be noted.

$$p^+ \longrightarrow n^0 + \beta^+ + \gamma \qquad {}^{52}_{25}Mn \longrightarrow {}^{52}_{24}Cr + \beta^+ + \gamma$$

Fig 8 Positron Decay Scheme

The positron thus created is then forcibly ejected from the nucleus in the same way as the negative beta particle previously mentioned. The fate of the positron is annihilation.

6. Positron Annihilation

The ultimate fate of a positron is a process known as annihilation. The positron, as a positive free charge, is unable to exist in the sea of negative charge (free electrons) which exists in all of nature.

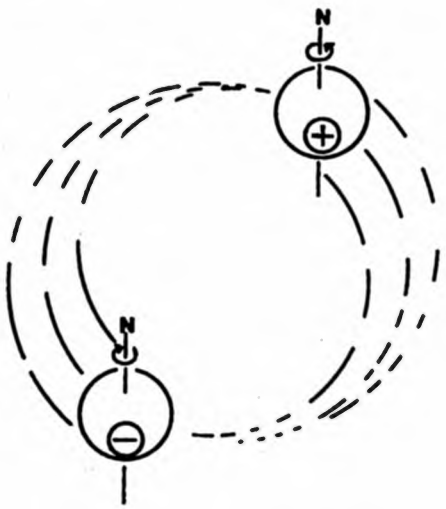


Fig. 9 Para-Positronium

The positron will lose its kinetic energy in much the same manner as an electron. When it has lost its kinetic energy, it then combines with a free electron in a fate known as annihilation.

Positronium is a quasi-element which exists just prior to the annihilation reaction. (Fig. 9) When the free positron has lost most of its kinetic energy, it combines for a fraction of a second with a free electron into one of two forms of positronium with a half-life which averages 1.25x10-10 seconds, for parapositronium, and 1.4 x 10-7 seconds for orthopositronium. If the spins of the positron and electron are in the opposite directions and the magnetic poles are parallel, the form is called parapositronium.

Parapositronium will disintegrate into two photons of .511 MeV each emitted in opposite directions (Fig. 10) These photons are representative of the rest mass energy of the two particles which have combined in annihilation.

Fig. 10 Positron Annihilation

In rare instances, however, if the spins are parallel and the poles are opposite, the form is called orthopositronium which will disintegrate into three photons, each of which acquires an energy of .340 MeV each (equivalent to 1/3 of the combined rest mass energy of the two particles).

If any energy is left in the positron prior to annihilation this energy will be given up as a low energy radiation beam. Capture of the positron by an electron is unlikely until the kinetic energy of the positron has been almost completely exhausted.

II RADIOACTIVE DECAY

A. Basic Principles of Decay

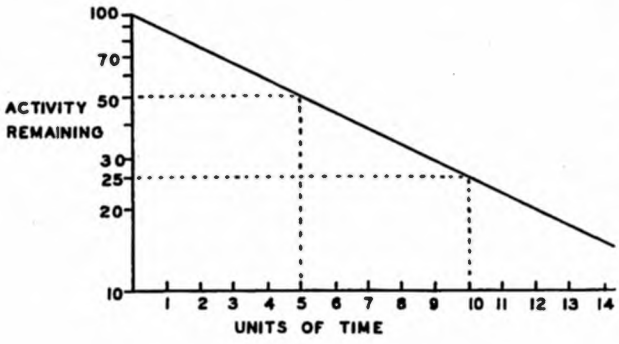


Fig. 11 Radioactive Decay

Radioactive decay is spontaneous and sporadic and is governed entirely by the laws of chance, however, due to the extremely large number of atoms involved it is very constant for each specific element. A radioactive nucleus may break down immediately, or it may never break down. On the basis of pure statistics, however, it has a certain chance of breakdown and, if enough atoms are present, a very definite half-life curve may be established on this basis. (Fig. 11).

The slope of the radioactive decay curve plotted on semi-log paper is always the same, provided only one radioactive element is present. Radioactivity is measured in curies. It should be noted, however, that radioactivity is measured by the indirect process of actually measuring radioactive decay. The amount of radioactivity in a particular substance is determined by the number of decays that take place per unit of time in this substance. By definition one curie of any material undergoes 3.7×10^{10} disintegrations per second. The rate of radioactive decay was originally discovered by Becquerel in 1896.

B. Utilization of Decay Factors

In practically every modern therapy installation in this country today there is at least one cobalt unit available. The theories of radioactive decay play an extremely important role in the calibration and dosage calculations of cobalt units.

One problem, however, that is inherent in the use of Cobalt 60 is its continuous decay. Decay necessitated frequent source strength computation for teletherapy units and interstitial sources. Ideally, one should compute these sources on a monthly basis. However, Cobalt's decay is approximately 1% per month and for this reason most centers are willing to compute decay for their sources every other month with the useful value being made to the mid-period. Fig. 12. (The actual decay of Cobalt 60 is 1.1023% per month.)

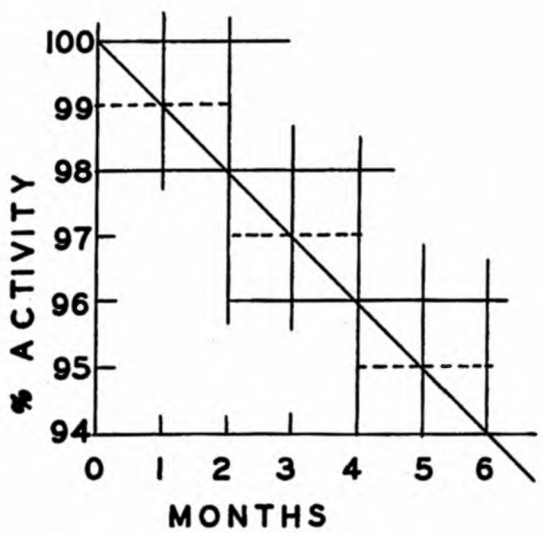


Fig. 12 Decay Calibration
—— Actual Decay - - - - - Applied Decay

This means that for the beginning of the first month the values used are 1% low and for the end of the second month 1% high. Most therapy patients will be treated at least a month to six weeks, therefore, this usually will average out to approximately the correct dosage during the course of the treatment. The same averaging will hold true for interstitial implants, provided the patient receives more than one implant separated by a reasonable period of time. If not, then individual decay computation should be done for the implant time. We have found it to our advantage to determine the present source strength or output by reference to the original amount rather than continuing application. This eliminates the possibility of compounded errors. Complete decay charts are available in the "Radiological Health Handbook" for the most commonly used isotopes.

The same principles hold true and may be used for the activity computation of Iridium 192 and Cesium 137. The only significant differences would be different decay factors, more frequent computation intervals for the Iridium, and more prolonged intervals for Cesium.

III RADIUM

A. History of Radium

Shortly after the discovery of x-ray by Röntgen in November of 1895, Henri Becquerel investigated the possible connection of x-ray screen fluorescence with natural mineral fluorescence. He began his experiments utilizing uranium compounds and accidentally noted that photographic plates were exposed by these chemicals. To Becquerel must go the honor of the discovery of radioactivity and his suggestion that this was very closely allied to the x-ray which Röntgen had only recently discovered.

Marie and Pierre Curie then began their series of chemical studies to extract and purify the radioactive material which was found in the uranium ore. They soon discovered that the activity in certain uranium ores was much higher than could be accounted for by the uranium content alone. It was their conclusion at that time that these ores must contain a new element. They isolated this substance in December of 1898, and named it radium.

Very shortly after its discovery, radium and radium compounds became the chief treatment for practically every known illness. As a result, over the next 10 to 20 years there were many tragic examples of radiation injury to both patients and physicians until the hazards of the indiscriminate uses of radium were eventually realized. It was found by a number of observers at that time that the use of these radioactive materials had a markedly deleterious effect on the growth of cancer. Indeed, it was found that cancer could even be cured by the judicious application of radium. From these early discoveries rose a complete new armamentarium for the treatment and cure of what had, up to this point, been practically an incurable condition.

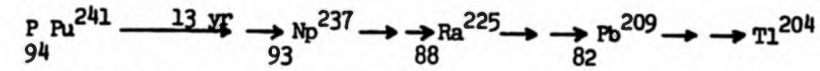
B. Radium Decay Scheme

There are four naturally radioactive series, one of which, the neptunium series, has died out due to the very short half-life of some of the elements involved. (Fig. 13). From these four series arise all of the naturally occurring radioactive materials that have been isolated. Each of these four radioactive series pass at one stage through an isotope of radium. Of these series, only one, the Uranium^{238} series, is of interest to the radiation therapist.

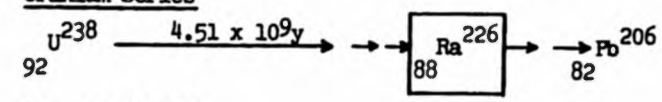
This series contains Radium 226, which is the isotope commonly utilized in radium therapy. The decay of the radium itself is of little consequence to us as the radiation (alpha particles) which arises from this particular step is not of therapeutic value. (Fig. 14). Radium decays to Radon 222 which very rapidly decays through a succession of steps eventually terminating in Lead 210. This rapid cascade of steps from Radon 222 through Lead 210 releases a variety of radiations. The two major steps in this scheme are the decays of Lead 213, (Radium B) and Bismuth 214, (Radium C). It is in these two major decay steps that the important gamma radiation is released.

The penetration abilities of both alpha particles and beta particles is very limited. In the usual utilization of radium the sources are sealed inside capsules whose walls usually are of sufficient density and thickness to stop all of the particulate radiations, but thin enough to transmit the majority of the high energy gamma radiation. The alpha particles are trapped within the

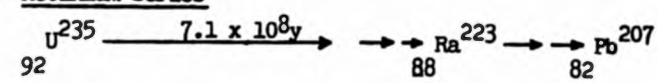
1. Neptunium Series



2. Uranium Series



3. Actinium Series



4. Thorium Series

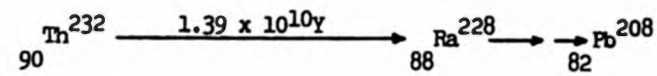


Fig. 13 The Four Naturally Radioactive Series

sealed source. The majority of the steps in the radium decay scheme are completely academic but are given in Fig. 14 to show complete continuity for interest. The important points to remember in the radium decay scheme are the particular steps that are responsible for the major gamma energies, as well as some of the pertinent half-lives.

C. Long Term Dosage from Radon

Radon, a gas, is easily separated from Radium 226, its parent. The separated radon gas is then intrapped within either small glass or gold containers referred to as radon seeds. These small containers of radon are then inserted by means of an appropriate implanting devise into tumors, allowed to pass through their principal gamma activity, and left within the body permanently.

The major gamma activity of Radon is insignificant after approximately one month's duration. (See Fig. 14). There remains, however, some gamma activity from Radon for many years due to bremsstrahlung arising from the energetic beta activity from Lead 210 (Radium D) and Bismuth 210 (Radium E). This activity may be measured with sensitive equipment as long as 20 years following an implant. It has been calculated according to Graham and reported in Johns, "The Physics of Radiology", that this contributes a dose in an average implant of from 100 to 300 rads over a 10 to 20 year period. It has been postulated that a chronic long term low dosage irradiation of this nature may have a carcinogenic effect.

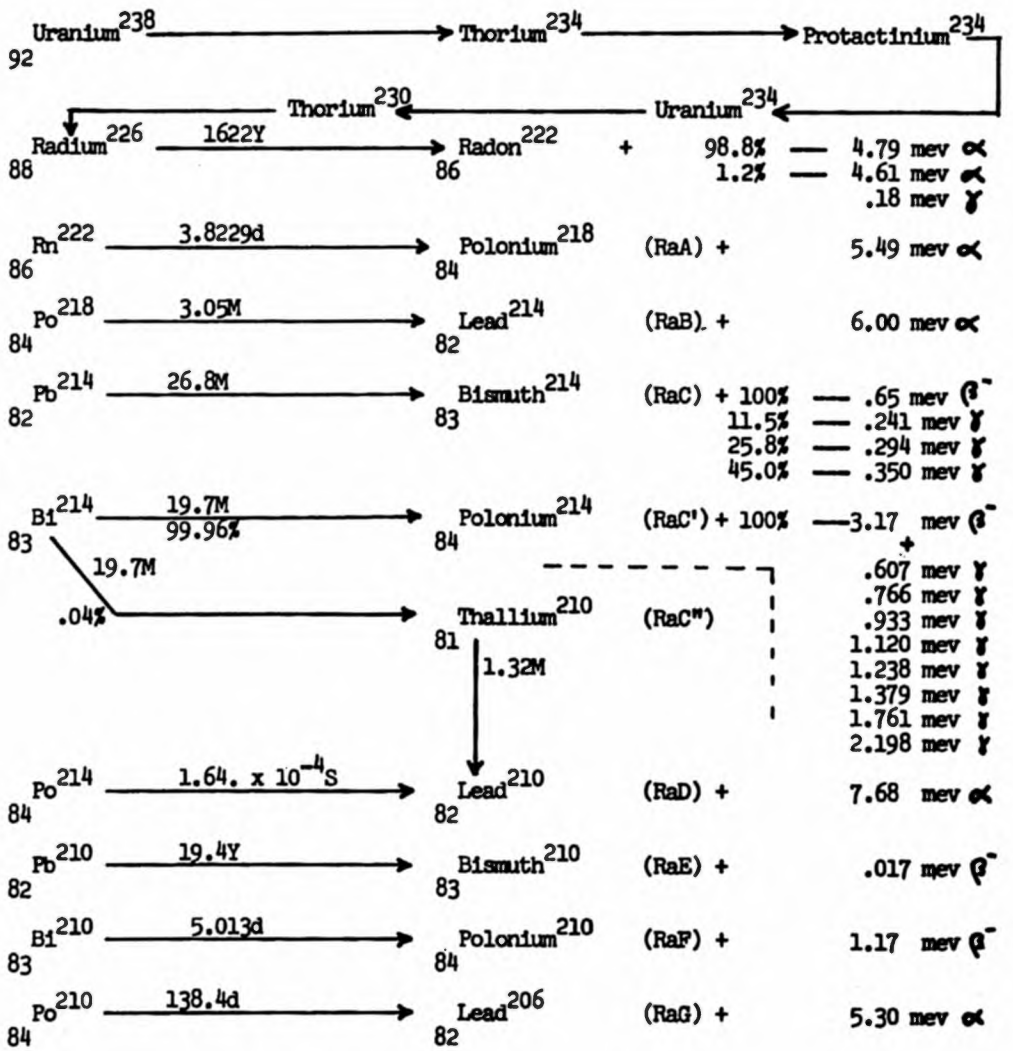


Fig. 14 Radium Decay Scheme

D. Leak Testing of Radium Sources

Since the very earliest application of radium sources the radon leakage from the incapsulated sources was recognized as a very distinct hazard. Pressure builds up within the sealed sources due to the continuous release of alpha particles which eventually become atoms of Helium gas. Over a period of many years a steady increase of pressure within these sources creates a tendency for them to rupture. Once this has occurred the radon gas is free to leak out, which it does as it is continuously produced. Numerous methods have been described utilizing various means of trapping and detecting the emanated radon gas.

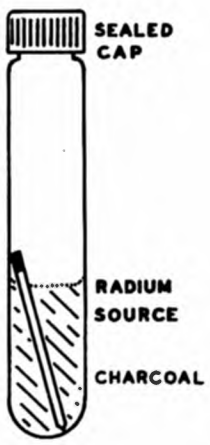


Fig. 15 Radon Leak Testing

A method which was developed by Dr. G.D. Adams, of the University of Oklahoma Medical Center, and recently published, has been found to be superior to all other methods of leak testing.

This method is quite simple and extremely adaptable. A test tube, which is capable of being sealed and capped, is partially filled with activated (Coconut) charcoal and the source under suspicion is inserted into the tube. The entire assembly is sealed and set aside in the radium safe overnight, or longer if greater sensitivity is desired. At the end of the exposure period, the test tube is carefully opened in a draft free environment, and the radium source is carefully extracted. The test tube is then promptly resealed. The determination from this point is quite simple. A rough determination may be made by holding the tube of charcoal with the entrapped radon near a geiger counter. Any indication on the geiger counter will indicate trapped radon within the tube and therefore a leaking source. A more sensitive method of carrying out an accurate analysis of radon leakage from the source may be made by counting the test tube with charcoal in a scintillation well counter of a known efficiency. It is then possible to determine the exact amount of radon leakage from the source under question.

If the radium tube is exposed to the charcoal for five hours or longer, then the fraction of saturation activity (meaning the maximum detectability for infinite exposure time) is found from the expression.

$$1-1.00781\ e^{-\lambda T}.$$

Where λ = decay constant for radon = 1.257×10^{-4} Min.$^{-1}$
and T = exposure time, in minutes.

Thus, if the combined factor for source geometry and detector sensitivity (well counter efficiency) is represented by S, and C counts per second above background are determined, the saturation disintegration rate would be written as the expression

$$C/S\ (1-1.00781e^{-\lambda T})$$

and the activity in pCi (Pico Curies) trapped on the charcoal would be found by the expression.

$$\frac{C}{.037\ S(1-1.00781e-\lambda T)}$$

A leaking radium source constitutes a very real and direct hazard to both the patient and the personnel utilizing such a leaking source. Such a source, without question, should be stored in isolation until such time as it may be returned to the manufacturer for either disposal or re-encapsulation. LEAKING RADIUM SOURCES SHOULD NEVER BE UTILIZED CLINICALLY. If the source was actually leaking radium powder or only the radon gas there may also be a question of contamination of the immediate working area and careful wipe testing of such areas should be carried out under the direction of your radiation safety officer.

E. Artificial Radioisotopes Used for Curie Therapy

A number of factors have been responsible for the replacement by many institutions of their supplies of radium with artificially produced radioisotopes. A number of objections to the use of radium have been made over the years, however, little could be done until the advent of readily available supplies of artificial radioisotopes in the early 1950's.

Chief among the objections to radium were its cost, approximately $20 per millicurie, and its contamination hazards. Many people have advocated the replacement of their radium sources with either Cobalt 60 or Cesium 137 sources. These sources may be made physically as small as radium and, in many cases, smaller. Cobalt is a relatively stable pure metal. The contamination problems for this isotope do not exist in the magnitude of those that exist with the use of radium since the decay products are non-gaseous. Cesium 137 is usually encapsulated as a cesium compound physically and chemically combined with a ceramic base material to prevent its accidental dispersement. Pure cesium metal is highly reactive chemically and, as such, would be very unsatisfactory to use. The chief disadvantage of the utilization of these materials for implantation work has arisen from the need of continuous decay correction due to their relatively short half-lives as compared to radium (See Fig. 16).

Isotope	R/Hr/mc @ 1 cm	γ Energy (MeV)	HVL (mm Lead)	Half Life
Au^{198}	2.5	.412	3mm	2.69 d
Co^{60}	13.4	1.173 1.332	12.mm	5.2 Y
Ra^{226} and equilibrium products (.5 MM PT Wall)	8.25	.053 ⟶ 2.43 (.55, 1.65) Equivalent energy	12 mm	1622 Y
Cs^{137}	3.3	.662	5.5 mm	30. Y
Ta^{182}	6.1	.05 ⟶ 1.24	9.8 mm	115 d
I^{131}	2.3	.080 ⟶ .722 (.364 87.2%)	3.1 mm	8.05 d
Rn^{222} and equilibrium products (.5 mm Pt Wall)	8.25	.053 ⟶ 2.43	12 mm	3.8229 d
Ir^{192}	5.5	.136 ⟶ .613 (.340) Equivalent Energy	2.5 mm	74.5 d

Fig. 16 Characteristics of Some of the More Commonly Used Radioisotopes. Adapted from Johns - "The Physics of Radiology" Second Ed, 1961. Courtesy of Charles C. Thomas, Publisher, Springfield, Illinois.

A number of other radioactive isotopes have been utilized as direct replacements for radium, each of which has its own advantages and disadvantages. Among these are Au^{198} and Ir^{192}.

Fig. 16 summarizes the pertinent information regarding the isotopes currently being used for intracavitary and interstitial implantation.

The final choice of the use of any of these materials as a radium substitute will depend entirely upon the personal preferences of the physician, the cost of the available isotopes, and the state of the available radium supply.

IV RADIUM DOSIMETRY

A. Conventional Radium Dosimetry

The early isolation of radium was in the form of the essential salts of radium, either as the chloride or sulfate. Because of the relative solubility of these materials it was soon realized that a method of containing them must be perfected in order to use their radiations safely. Encapsulation of the radium source within either platinum, gold, or silver tubes was utilized very early as a method of containing the highly dispersible salts. Encapsulation also serves to selectively filter out the high energy undesirable particulate radiation (Alpha and beta particles).

An arbitrary unit of measure of the amount of radium present was decided upon. The activity of one gram of radium was originally designated as 1 curie. It therefore followed that 1 milligram of radium was 1 millicurie. One curie was later established to be 3.7×10^{10} atoms disintegrating per second. The activity of one milligram of radium is 3.7×10^{7} disintegrations per second. One millicurie of radon is the amount of radon in equilibrium with 1 milligram of radium. Approximately 1 month of time is required for the equilibrium of radium with its important daughter products in a fresh container.

Early radium dosimetry was carried out in the form of milligrams administered for a period of hours, or, as commonly expressed in milligram hours. It should be noted that due, to the extremely high chemical activity of radium, it is never kept in a pure state but is usually combined with chlorine in the form of radium chloride. The statement that a given sample of radium contains 100 mgm of radium actually means that there is a total weight of 131 mgm of radium chloride, 100 mgm of which would be pure radium element.

Radium has been used for many years in the treatment of certain types of tumors. It has been found to be extremely useful in treating tumors which are in accessible locations such as skin, oral cavity, uterus, and cervix. In these examples the radium may be inserted into the area to be treated and left there for a specified length of time while delivering a satisfactory dose of radiation to the area under treatment. The use of radium implantation in conjunction with external beam therapy is very common. The usual form of treatment for carcinoma of the cervix is considered to be a combination of cervical radium applications for high intensity irradiation of the central lesion, combined with external beam therapy for prophylatic irradiation of lateral tumor and pelvic wall or lymph

node involvement. In the early years of radium dosimetry, dosage delivered to the tumor was expressed in milligram-hours. Statements of dose in this unit have a meaning only when the geometric arrangement of the sources is reproduced exactly from one patient to the next. Exact geometrical reproduction from one patient to the next is quite difficult and, in most cases (with the exception of some gynecological work) an expression of dosage in milligram-hours is only a rough indication of the total dosage delivered.

If one wishes to calculate the radiation contribution to the cervix and other points throughout the pelvis from the radium application, this may be done, but with considerable difficulty and time consumption. For many years rough calculations have been carried out for a few specific points within the pelvis. The radium sources are usually considered as point sources and simple calculations are made to determine approximate dosage at these various levels. These doses are added appropriately to the external beam therapy doses for total treatment planning.

In the early 1930's Paterson and Parker devised a system of radium dosage called the Manchester System which enabled the radiotherapist to express the dosage to the patient in roentgens. By the utilization of the Paterson-Parker dosage table the physician is able to decide upon the dose to the tumor in rads for a given implant. There are certain very specific rules, however, which must be exactly followed in the Paterson-Parker system or the calculations will be invalid.

The Paterson-Parker data were calculated by means of Sievert Isodose curves considering each source as a linear source. The system used by Quimby in the construction of her tables, considered each source to be made of a number of point sources, and the dosage contribution from each point to the area in question was calculated. Both systems assume the dosage at 1 cm from 1 mm radium filtered by .5 mm Pt to be 8.4 R/hr. These values must be corrected by .983 to correspond to the present accepted value of 8.25 R/hr.

B. Basic Computer Theory

Modern electronic computers may be broken down into three major categories: Analog, digital, and hybrid.

1. Analog

Analog computers may be compared to the common slide rule in which an approximation of the number, the accuracy of which depends upon the need of the problem, is established analogous to a mechanical position, or in the computer, a voltage. The mechanical positions or voltages of the problem are then compared and an answer is read as either a third mechanical position or voltage. The analog computer is extremely fast, the answer being established as soon as the parameters of the problem are completed. There are many instances in which analog computers are superior to digital units, although the inherent design does not allow the accuracy of a digital unit.

2. Digital Computers

The digital computer is just as stated. Numbers are used in this instrument in a one to one ratio. In other words, one number is equal to one electrical or mechanical operation within the machine. Common examples of digital computers are the abacus, the adding machine and the desk calculator. The complete numbers to be evaluated are used, and all operations, including division and multiplication, are carried out as either addition or subtraction problems. Digital computers may be considered decimal, usually to the base 10, or binary utilizing the base 2.

Binary computers are better suited for scientific problems with large amounts of computations and relatively small amounts of output information. For data processing problems with a larger amount of input and output information and relatively few computations the digital system based on the decimal of 10 is a much better arrangement, although quite often the computer may handle the numbers internally in binary form. Many installations may use two computers with the second computer designed primarily to digest information on the input and output sides of a problem to feed the faster main computer for the problem solving operations.

Digital computers accept numbers, letters, and symbols. Information is usually fed into the system from punched cards, punched paper tape, magnetic tape, or inserted manually from a keyboard or switches. The computer must operate under the direction of a control unit. The sequence of steps to be performed must be translated into detailed information which the computer can understand.

A series of instructions is called a program. When it is retained in a storage device, it is called a storage program. These coded instructions in storage are available to the control unit as needed to direct and complete an entire sequence of operations. Special instructions may enable the logical arithmetic unit to make decisions based on intermediate results; these decisions allow the computer to select the proper course among several alternatives for solving the problem.

Today's high speed electronic digital computers can handle alphabetic data as well as numeric data. Computers are capable of performing complicated calculations, manipulation of information, and can even make logical decisions based on information available to the machine. All of these calculations are carried out at fantastic speed, and the machines are becoming smaller and even more efficient due to the use of completely solid state transistorized circuitry.

Information and data may be internally stored within the computer by electro-mechanical, magnetic, or electric devices until it is needed. This information is stored in a manner quite similar to the way in which music or speech is stored on a home tape recorder for playback. Stored information is accessible and may be referred to once or many times, and may be replaced or changed if desired.

The IBM 707 computer has 100,000 positions of data storage in its internal memory. This is information that is readily available to the program being operated at that time. Almost an unlimited amount of information may be stored externaly in the form of magnetic disk files or tapes. This information is re -

ferred to as the library and may be made available to the computer on command to be stored within its internal memory for use in solving the problem at hand. One of the limiting factors in the speed of a computer is the accessibility of the stored information to the main control unit. This is referred to as access time or the amount of time required to obtain information from a storage position and make it available to the other units of the computer system. The heart of the computer is the logical arithmetic unit which is capable of addition, subtraction, multiplication, division and comparison of numbers all at fantastic speeds.

After the computer has completed its work it produces answers in many different forms. The results may be punched into cards, recorded on magnetic tape or printed out on a report sheet. Printers provide a high speed computer output by printing an entire line of information at one time.

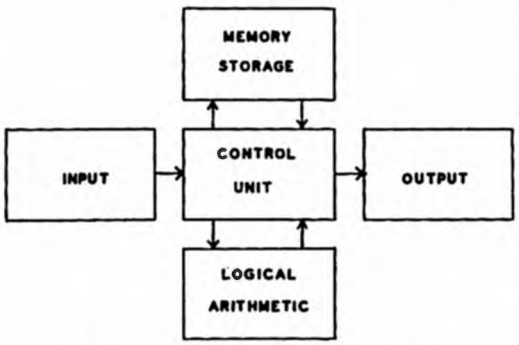


Fig. 17 Organization of the Basic Components of a Computer

The functioning of the elements of the computer may be compared to the steps required for solving a problem by pencil and paper methods. The input would correspond to the information given in the problem. A knowledge of arithmetic controls the handling of the problem. The logical arithmetic unit performs the same function as manual calculations. Storage may be compared to the scratch papers on which intermediate answers are noted. The answers to the problem would then be considered as the output.

The number of instructions required for the complete solution of a problem may be a few hundred or many thousands, depending upon the problem. The computer refers to them one after another, or can be instructed to repeat, modify or skip over certain instructions. These operations depend upon the intermediate results or circumstances. The computer, however, must be able to recognize such circumstances and anticipate them with appropriate instructions being included in the program.

3. Hybrid Computers

A third type of computer is called the Hybrid computer or a combination of digital and analog systems. This allows various pieces of information to be handled in either an approximate high speed fashion or in an extremely exact but slower speed fashion depending upon needs.

It should be realized that computers are only as "smart" as the people who have written the program being used. A computer can do only what it is told to do and it will do exactly what it is told, right or wrong, at the speed of approximately 90,000 computations per minute.

C. The Utilization of Computers in the Calculation of Radium Dosage

The advantages of intracavitary and interstitial radium therapy have long been acknowledged. The chief difficulty in the technique of dose distribution around radium implants is to be certain that the patient receives adequate therapy but, at the same time, not to exceed the limits of tolerance of existing normal structures.

In the attempt to solve the problem posed by these difficult dosage determinations, and the dissatisfaction with less than adequate determinations, a number of computer programs for estimation of dosage around implants have been constructed.

The basis for most of the existing computer programs consists of obtaining either a pair of stereo-radiographs, or a true AP and lateral film, of the radium application. In either case an adequate magnification factor measurement must also be included with these films. We use a disc of radio-opaque plastic (5 cm. in diameter) placed at the level of the implant for all films. A disc is used because a disc, ring, or sphere always projects a true diameter no matter at what angle it is viewed.

To attempt to go into the various complexities of computer program setups for radium calculations is far beyond the scope of this work. The various radium computer programs presently being utilized are almost as varied as the institutions utilizing them. A superficial understanding of the basic theory of computers, and a good basic understanding of the problems of radium dosimetry are really all that is required of the radiation therapist in these programs. The actual construction and operation of an adequate radium program may then be carried out by the personnel of your computer center, who ultimately have the responsibility of writing the computer programs.

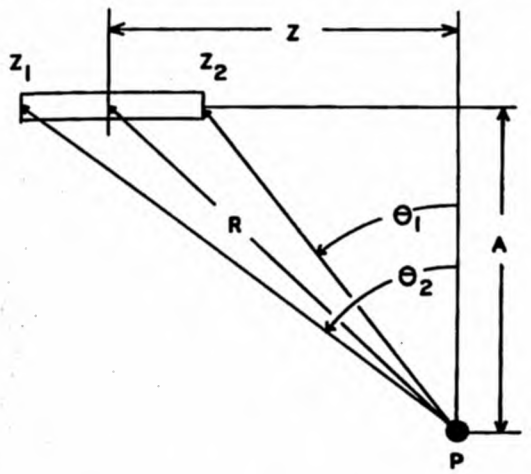


Fig. 18 Seivert Integral

The geometrical information for the program input is usually obtained from the films. The information necessary to be extracted from the films is directly dependent upon the amount of information available as well as the total accuracy desired. In most cases careful measurements are made of the angular displacements of the radium sources and co-ordinates are then taken in relationship to the planes of the films. These measurements enable the computer to re-construct the radium implant within its own memory.

Two basic types of programs at the present time are being employed, these are calculation, and table look-up.

Calculation type programs, as a rule, utilize the Sievert integral. The Sievert integral type of calculation enables the computer to take any number of previously selected points in the proximity of the implant and calculate the exact radiation contribution to these points from each of the sources of the implant. These calculations are quite involved and generally require the dosage contribution from a minimum of 3 points along the linear radium sources, (both ends and the central portion.) These programs also take into account the difference in filtration due to the angulation of the sources and many steps are required for the exact answer. (Fig. 18)

$$I = \frac{Kp}{A} \int_{\theta_1}^{\theta_2} . \, d\theta . e^{-ud/\cos \theta}$$

When the thickness of the platinum filter (d) is small, the variations of the filtration correction with angles may be neglected and this equation may be re-written.

$$I = \frac{Kp}{A} \, (\theta 2 - \theta 1) \, e^{-ud}$$

K = Emmission Constant for unfiltered Radium (8.81 Rads per Mgm-hr)
p = Linear density of the radium
d = Millimeters of platinum wall filtration
P = (Fig. 18) point of interest
u = Linear absorption coefficient of the filter.
I = Dose rate from the entire source
A = (Fig. 18) Right angle distance from the parallel axis with the source to the point of interest.

Sievert integral calculations adapted from Johns "The Physics of Radiology" Second Ed. 1961, Charles C. Thomas Publisher, Springfield, Illinois.

The table look-up type of radium calculation is far simpler than the Sievert integral type but for the marked increase in simplicity one must sacrifice some slight accuracy and have a larger computer memory available. The table look-up system considers the radium sources individually and reconstructs right angle triangles to the specific point of interest. The computer's memory contains a massive radium dosimetry table, as well as the mathematical possibility of extrapolating between numbers of the table. The utilization of this system simply allows the computer to look up in its memory and sum the dosage output from each source to each point.

D. Computer Program Information Display

The information display or the eventual writeout of the dosimetry information may be presented in as wide a variety of methods as there are computer programs. In Fig. 19, to the left, an isodose pattern of radiation is displayed by the utilization of symbols corresponding to a dosage code which is included with each sheet. The other method, (Fig. 19), above, utilizes a slightly different approach by having the computer actually print out the numerical values for the entire plane. These values may then be connected by lines and the isodose curves determined from this. Either method works quite satisfactorily and the choice depends upon personal preference and the capabilities of the computer.

PLANE 3

IDENTIFICATION NUMBER- 64-6191 DOSES RATE CALCULATED FOR 3/18/64

DOSE RATE AT ALL LATTICE POINTS MEASURED IN ROENTGENS PER HOUR

Y \ X	50	45	40	35	30	25	20	15	10	5	0	-5	-10	-15	-20	-25	-30	-35	-40	-45	-50
-50 MM	1	1	1	1	2	2	2	3	3	3	3	3	3	3	2	2	2	1	1	1	1
-45 MM	1	1	1	2	2	3	3	3	3	3	3	3	3	3	3	3	2	2	1	1	1
-40 MM	1	1	2	2	3	3	3	4	4	4	4	4	4	4	3	3	3	2	2	1	1
-35 MM	1	2	2	3	3	4	4	5	6	6	6	6	6	5	4	4	3	3	2	2	1
-30 MM	2	2	3	3	4	4	5	6	7	8	8	8	7	6	6	4	4	3	3	2	2
-25 MM	2	3	3	4	4	6	7	9	10	12	12	12	10	9	7	6	4	4	3	3	2
-20 MM	2	3	3	4	6	7	9	12	15	18	19	18	15	12	9	7	6	4	3	3	2
-15 MM	3	3	4	5	6	9	12	17	24	30	34	30	24	17	12	9	6	5	4	3	3
-10 MM	3	3	4	6	7	10	15	24	37	58	72	58	37	24	15	10	7	6	4	3	3
-5 MM	3	3	4	6	8	12	18	30	58	129	221	129	58	30	18	12	8	6	4	3	3
0 MM	3	3	4	6	8	12	19	34	71	219	0	219	71	34	19	12	8	6	4	3	3
5 MM	3	3	4	6	8	12	18	30	58	129	221	129	58	30	18	12	8	6	4	3	3
10 MM	3	3	4	6	7	10	15	24	37	58	72	58	37	24	15	10	7	6	4	3	3
15 MM	3	3	4	5	6	9	12	17	24	30	34	30	24	17	12	9	6	5	4	3	3
20 MM	2	3	3	4	6	7	9	12	15	18	19	18	15	12	9	7	6	4	3	3	2
25 MM	2	3	3	4	4	6	7	9	10	12	12	12	10	9	7	6	4	4	3	3	2
30 MM	2	2	3	3	4	4	6	6	7	8	8	8	7	6	6	4	4	3	3	2	2
35 MM	1	2	2	3	3	4	4	5	6	6	6	6	6	5	4	4	3	3	2	2	1
40 MM	1	1	2	2	3	3	3	4	4	4	4	4	4	4	3	3	3	2	2	1	1
45 MM	1	1	1	2	2	3	3	3	3	3	3	3	3	3	3	3	2	2	1	1	1
50 MM	1	1	1	1	2	2	2	3	3	3	3	3	3	3	2	2	2	1	1	1	1

PIERCING POINTS FOR THE RADIOACTIVE PORTION THROUGH PLANE 3 IZ-COORDINATE = 0.0MM
NEEDLE NO. X-COORDINATE Y-COORDINATE
1 .00 -3.0

04 TANDEM SOURCES. 20.00. 22.50. 20.00. 22.50 MG
04 OVOID SOURCES. 20.00. 20.00. 20.00. 20.00 MG

37 30 92 BROWN. MARTHA 20APR66 4=INTAN NO. PROGRAM 25APR66. HOW=2I2I2I2I2I2D

PLANE Z=+0.0 CM. NORMAL SCALE. A=1000. B=2000. C=3000. ETC TO I=9000 RADS
8 RIGHT 6 5 4 3 2 -1 0 +1 2 3 4 5 6 LEFT

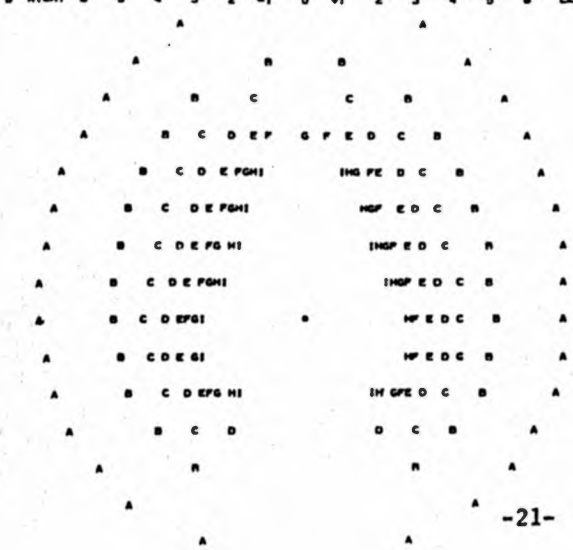


Fig. 19 - Above, typical computer printout in use at the Mallinckrodt Institute of Radiology, St. Louis, Missouri. The numbers represent dosage delivered to a grid pattern usually calculated on a number of planes throughout the pelvis.

Left, Computer printout currently in use at the University of Oklahoma Medical Center, Oklahoma City, Oklahoma. Letter codes are used to represent dosage levels.

V X-RAY PRODUCTION

X-rays are produced when high speed electrons are brought to rest in a target material. In the modern x-ray tube, used in both diagnostic and therapeutic radiology, the electron beam is generated by emission from a hot wire filament by thermionic effect. These electrons are then accelerated by means of an appropriate kilovoltage through the evacuated tube, finally striking the target and generating a stream of x-rays.

Once the overall picture of the high-speed electrons achieving a sudden deceleration within the structural material of the target is in mind, one must take a deeper look down to the atomic level, to understand more fully the actual occurrences within the matter of the target itself.

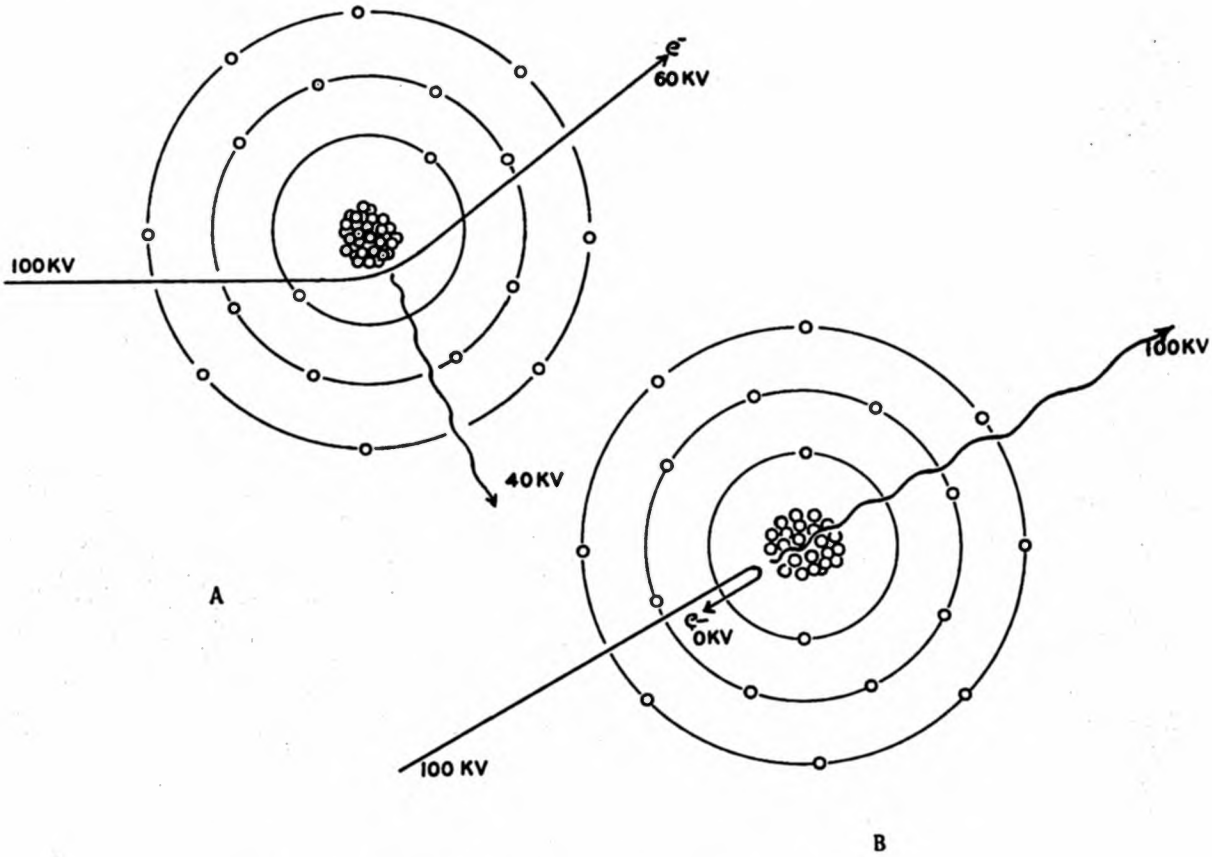


Fig. 20 Production of Bremsstrahlung Radiation

A. Bremsstrahlung Radiation

Upon approaching the strong electrostatic fields of the nucleus of the target atom, the electron may undergo an abrupt slowing down or deceleration. (Fig. 20-A)

The loss of kinetic energy associated with this deceleration is converted to an x-ray photon of equivalent energy. The term "Bremsstrahlung", or braking radiation, has been applied to this process. Note that in the example, the electron approached the atom at an energy of 100 kilovolts. Deviation occurred as the electron passed through the electrostatic field of the nucleus with a subsequent deceleration throughout the curving portion of its path. It left the vicinity of the target atom with only 75 kilovolts of energy, the 25 KeV which was lost was emitted as a single photon of x-ray which leaves at an angle to the deviation path, this angle being determined by sidewise momentum given to the electron.

Bremsstrahlung has often been referred to as continuous or white radiation since, like white light, it consists of a continuous range of wave lengths. This process of stopping a particular electron may result in an energy loss by the electron of any fraction of its original energy from the smallest amount to its maximum energy. This energy loss gives rise to a photon of x-ray of corresponding energy. A complete loss of the total energy of the electron (Fig. 1-B) occasionally occurs when an electron approaches the nucleus head-on and is completely stopped by the nuclear electrostatic field. In this example, the electron entering the target atom has an energy of 100 kilovolts. The actual occurrence that takes place here is a deviation such as was seen in Fig. 1-A, however, the reflection of the electron is completely back upon itself with a total loss of energy at the point of 180° reflection. This gives the generated photon an energy which is equivalent to the energy of the input electron, in this case 100 KeV. The electron then drifts out of the vicinity of the atom with zero amount of energy. This gives rise to the minimum wave length in the continuous spectrum.

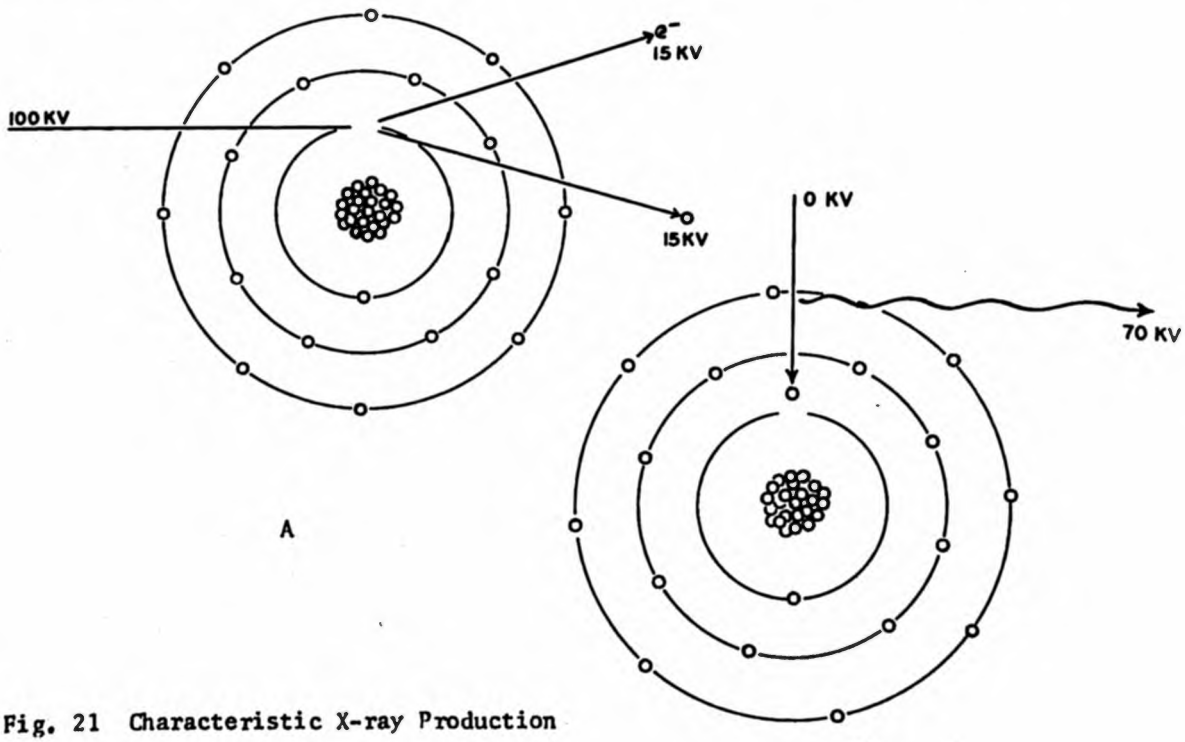


Fig. 21 Characteristic X-ray Production

B. Characteristic X-rays

An electron with sufficient minimum kinetic energy may appear to collide with an inner orbital electron of a target atom and eject this electron from its orbit. To free this electron, work must be done against the attractive forces of the nucleus.

For example: In Tungsten 69.51 Kilovolts (Diagrammatically 70 KeV) is required to eject the "K" shell electron from the atom. (Fig. 21-A)

The atom is now unstable, or ionized. Immediately, the orbit vacated by the electron is filled by another electron dropping into it from one of the outer orbits, or in some cases by a free electron. (Fig. 21-B) Since energy was required to dislodge the "K" electron initially, this amount was absorbed from the impinging electron. When an electron then fills this area, a like amount of energy must be emitted. (Satisfying the law of conservation of energy). This energy is emitted in the form of x-ray called "Characteristic radiation", because it has a discrete wave length characteristic of the target element and the orbits involved. Often several electrons in cascade from the outer shells emit a number of photons as vacated shells are filled - however, the total energy released must add up to the energy of the orbit originally vacated.

VI INTERACTION OF X-RAY WITH MATTER

When an impinging beam of radiation strikes an absorber three things are possible:

1. A portion of the beam of x-ray may be transmitted.
2. A significant portion of the beam may be absorbed.
3. A fraction of the beam may be scattered, that is, it exits from the material with a direction change and often an energy modification from its original state.

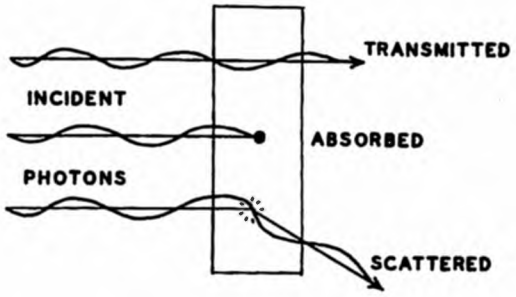


Fig. 22 Interaction of X-ray with Matter

In radio-diagnosis a major portion of the primary beam is absorbed selectively in various materials, bone, tissue, and fat, some is scattered, and the remainder of the beam is transmitted to give the image upon the film. The scattered component is frequently minimized by grids or other mechanical devices.

The energy deposited in tissue from the components which are absorbed directly or scattered constitutes the dose delivered and thus is of great interest in radiation therapy.

The fraction of the beam which is transmitted is of no concern, except as an efficiency loss.

The scattered component of the x-ray beam is also of considerable importance. Energies of the scattered components are directly dependent upon the energy of the primary beam and their characteristics will be taken up in greater detail.

High speed electrons are usually produced as a product of the first interaction of the photon and the atoms of matter. These high speed electrons actually act as the chief transfer agent of the energy of the primary photon beam to the matter itself. These electrons lose their energy to matter by the following methods: See Fig. 32

1. Ionization
2. Excitation
3. Breaking of molecular bonds
4. Bremsstrahlung, which gives rise to secondary photons
5. Heat, the ultimate degradation product of all of these processes.

The chemical effects of heat, molecular bond rupture, excitation, and ionization are the chief causes of biological changes which are observed following radiation therapy.

In biological systems, the primary photons create considerable high energy secondary electrons. The secondary high energy electrons thus created have considerable energy and interact very readily creating further photons which create further electrons. This process may last for as many as 30 cycles in the average biological system before the energy expenditure has been sufficient to terminate the interaction.

In order to interact with electromagnetic waves the object of interaction, as a rule, should have roughly the same size as the wave length of the impinging electromagnetic wave.

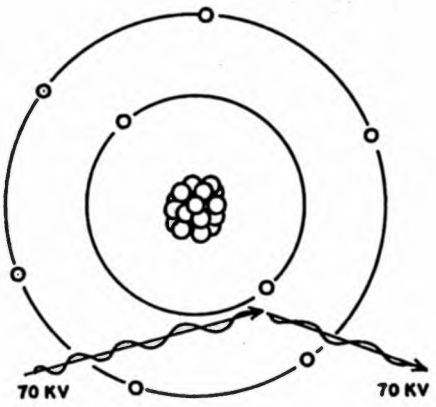
1. Low energy electromagnetic waves, that is, those waves with relatively long wave lengths ($\lambda \cong 12.4$Å), on the average of one kilovolt. These waves will interact with the entire atom.

2. High energy waves, that is, those with a somewhat shorter wave length ($\lambda \cong .124$ Å), on the order of 100 kilovolts are capable of interacting with electron shells. Most diagnostic x-rays fall within this range.

3. Very high energy waves with extremely short wave lengths ($\lambda \cong .00124$Å), 10 MeV or higher. These waves tend to interact with the nuclei of the atoms. This is not as much due to size as to the extremely high energy concentrations which are found within the nucleus.

A. Thompson or Unmodified Scatter

In Thompson or unmodified scatter the photon passes in the immediate vicinity of the atom, however, the only interaction that occurs is a change in direction without any measurable energy loss. This is also known as classical or coherent scattering. It is postulated that the electromagnetic wave passing near an



electron sets it momentarily into vibration which causes it to radiate energy at the same frequency as the incident electromagnetic wave. A variation on this theory holds that the photon is completely absorbed by the atom which is then set into vibration and re-emits a photon of exactly the same energy as that which was absorbed, as a rule the direction of the photon will be changed. This process is of very little interest to radiation therapy, as no energy is absorbed into the tissue involved. This process is of great importance, however, in dealing with diffraction x-ray spectrometry.

Fig. 23 Thompson Scatter

B. Photoelectric Effect

Classically photoelectric effect is considered as the interaction of a photon with one of the orbital shell electrons (Fig. 24). This effect usually occurs with the tightly bound electron of one of the inner shells. Almost always this will be the "K" shell electron, if the photon energy is at least as great as the binding energy of the "K" shell electron, occasionally the "L" shell and, very rarely, the "M" shell. This process is often confused with characteristic x-ray production (V-B) as characteristic x-ray production is a secondary effect of photoelectric collision. In photoelectric collision the photon is completely absorbed. The orbital electron which was struck by the photon then leaves the nucleus with an energy equivalent to the energy of the impinging photon minus the electron binding energy of the shell. The example given shows the "K" shell binding energy for oxygen which is approximately 500 electron volts. The electron which is ejected is termed the photoelectron (Fig. 24). The vacated orbital shell is then filled with either a free electron or perhaps an electron from one of the adjacent shells (Fig. 25).

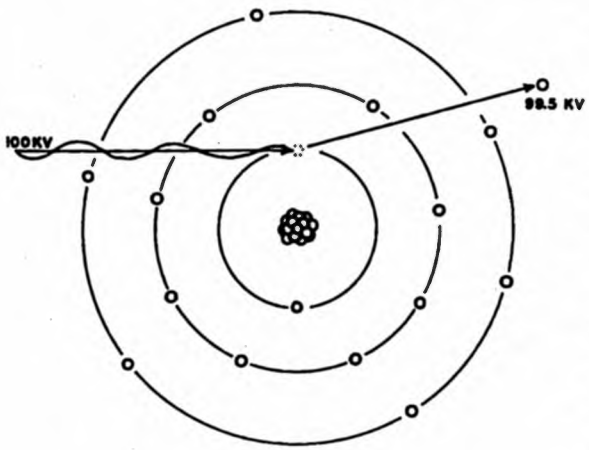


Fig. 24 Photoelectric Effect. Photon Interaction with a Low Atomic Weight atom.

Note that as the vacated shell is filled, a photon is released which is of very low energy and most characteristically will interact with other electrons within the immediate sphere of the same atom. The subsequent release of energy

is characteristic of the shells which have been alternately vacated and filled with additional electrons, this is termed an Auger electron shower. (Fig. 25)

The average binding energy of the "K" shell for tissue is approximately 500 electron volts. (This is the value for oxygen). Due to this low energy, the photoelectron acquires practically all of the energy of the photon. The extremely low energy of the characteristic photon released within the tissue is responsible for the Auger effect which takes place with the eventual total dissipation of energy within the original atom.

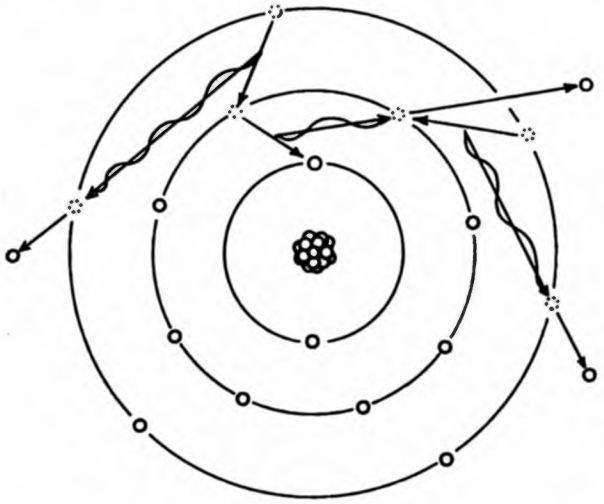


Fig. 25 Characteristic Radiation Emmission from Photoelectric Interaction in Tissue.

The photo-electron created continues through the tissue creating further excitation, ionization, bond rupture, and other interactions.

As the atomic number (Z) increases the chance of photoelectric collision increases per gram of material as the 3rd power (Z^3). Photoelectric collision is generally considered in terms of the entire atom. High atomic number elements have the same number of "K" shell electrons (2) as low atomic number elements, however, they have a smaller radius of the shell and therefore a higher concentration of electrons per unit area. The total number of electrons present is therefore, also a significant factor (Z^1), giving the total probability of photoelectric interaction per atom as Z to the 4th power (Z^4).

As the energy is increased the probability of photoelectric interaction occuring decreases. With a rise in effective KeV one sees a decrease after an initial peak in the photoelectric effect. Peaks are noted from the various shells of interaction. We may conclude that the more tightly bound electrons are the more important ones in bringing about photoelectric absorption, and the maximum absorption occurs when the photon has just enough energy to eject the bound electron. A resonance effect with the binding energy and the energy of the photon occurs at a point where the energy of both is extremely close. This is the point of maximum absorption of beam energy. One can easily see that there is a very close relationship between the energy of the impinging beam of photons and the atomic number of the material under study.

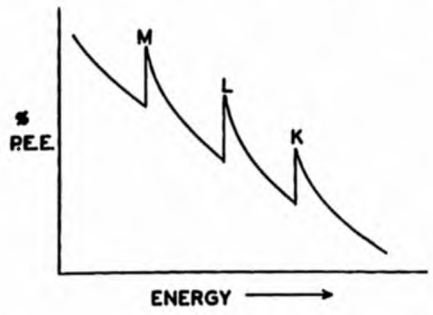


Fig. 26 Photoelectric effect as related to energy

C. Compton Effect

Compton effect is equally likely to occur with any electron, more usually with the more numerous electrons in the outer shells, and at higher energies then photoelectric effect. Compton is significant in the range above 100 kilovolts. (Fig. 30). The loosely bound outer orbital electrons are almost free electrons, that is, they have extremely low binding energy. The probability of Compton interaction per atom depends upon the number of electrons present. (Z) In the neutral atom, this number is exactly the atomic number of the material. Compton effect is dependent upon the atomic number or "Z" only. Compton effect is independent of atomic mass. The amount of interaction that takes place and the amount of energy exchanged by this interaction depends on the energy of the incident photon and on the angle through which it is scattered.

The original photon is scattered and the electron of the orbital shell is ejected. (Fig. 27). The electron is then called the Compton or the recoil electron. The energy transfer depends entirely upon the angle of hit. In a direct hit the electron is ejected forward and the photon recoils backward 180 degrees on its original path with minimal energy left. (Fig. 28).

This can be likened to playing pool in which you may have the entire range between a direct hit by the cue ball on the target ball to a mere glancing blow on the target ball with very little energy loss by the cue ball. The wider the angle of reflection θ becomes, the more energy is transferred to the electron and the lower the energy remaining in the scattered photon (Fig. 28).

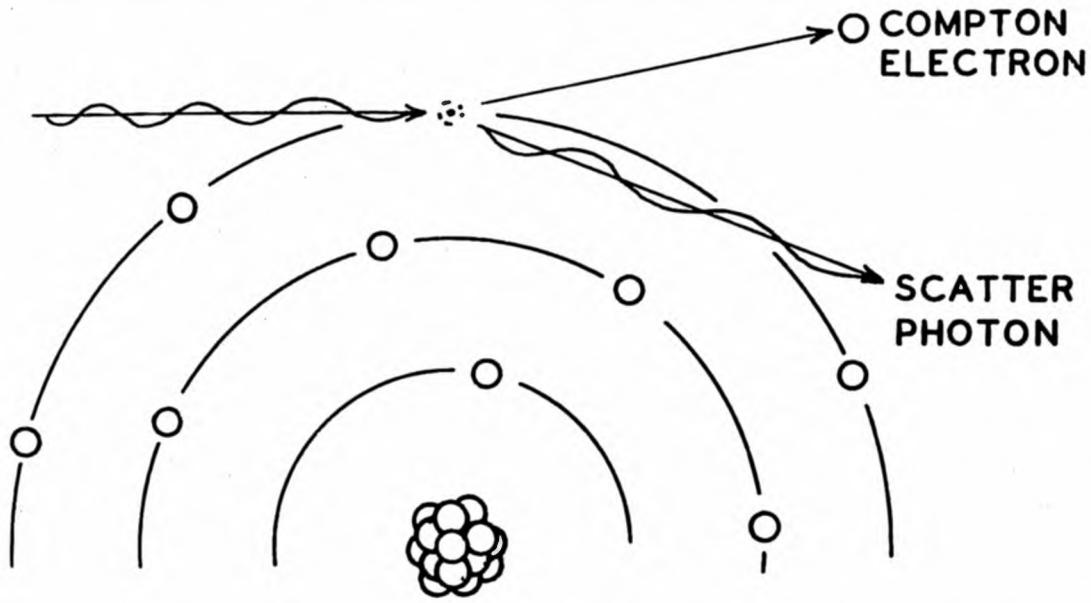
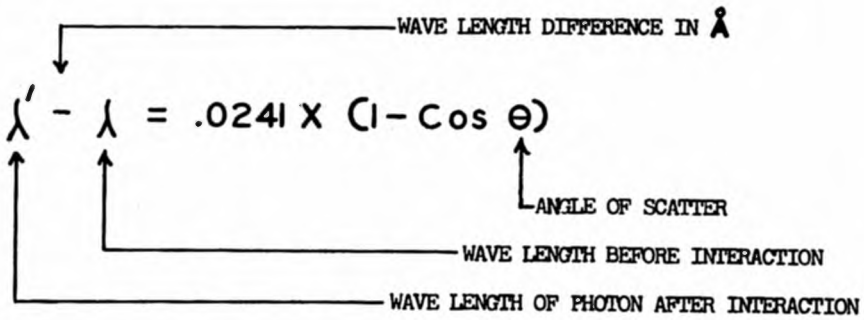


Fig. 27 Compton Effect. Adapted from Johns "The Physics of Radiology" Second Ed. 1961, Charles C. Thomas, Publisher, Springfield, Illinois.


WAVE LENGTH DIFFERENCE IN Å
λ' - λ = .0241 X (1 - Cos θ)
ANGLE OF SCATTER
WAVE LENGTH BEFORE INTERACTION
WAVE LENGTH OF PHOTON AFTER INTERACTION

The cosine of the angle varies between +1 and -1. Minus 1 is when the angle is equal to 180 degrees (Fig. 28-B). The parenthetical expression gives 1 minus -1 or +2. With this particular arrangement one will see the maximum energy loss of the incident photon and the maximum energy gain of the recoil or Compton electron.

Minimal energy loss of the incident photon occurs when the Cosine is almost equal to +1 or the angle is nearly zero (Fig. 28-C) and the energy loss is zero. One minus a +1 is equal to zero. In actuality no observable interaction occurs and no energy is transferred in this process.

Fig. 28-A Angle of hit - Compton interaction

Fig. 28-B Maximum angle of hit Cos. θ = -1

Fig. 28-C Minimum angle of hit Cos. $\theta \cong +1$

The amount of energy given to the recoil electron and the amount of energy retained by the scattered photon is dependent upon the energy of the original photon. In a collision involving a lower energy photon the scattered photon has nearly the same energy as the primary impinging photon and the recoil electron acquires only a small amount of energy. A single Compton scattering event at low energy does not result in an appreciable amount of energy being transferred to the electron and, hence, to the absorbing system. However, multiple compton scatterings along the history of a given photon can lead to appreciable energy transfer to the system.

In instances where the incident photon is of high energy, the energy of the scattered photon is usually much less than the primary; the recoil electron then acquires most of the energy of the primary photon. A high energy incident photon involved in a Compton absorption gives rise to a large amount of energy in the form of electron tracks which can produce considerable biological damage. Compton absorption is more important than the other forms of photon absorption in the range from 100 KeV up to 10 MeV. (See Figure 30). This is especially true when soft tissue is involved, however, as previously noted Compton absorption is governed by the total number of electrons in the material and not so much by the atomic number or the atomic weight. Bone and tissue appear quite similar at high kilovoltage because the majority of the interaction in this range is Compton. This is the principle reason that cobalt irradiation and betatron irradiation can treat tumors in and around bone almost as successfully as though there were no bone present. This is also the reason that radiographs taken at high energies, utilizing super voltage machines, fail to give a sharp distinction between soft tissue and bone, such as is seen in the lower diagnostic energies where photoelectric collision plays an important effect and absorption in bone is much higher (See Fig. 37)

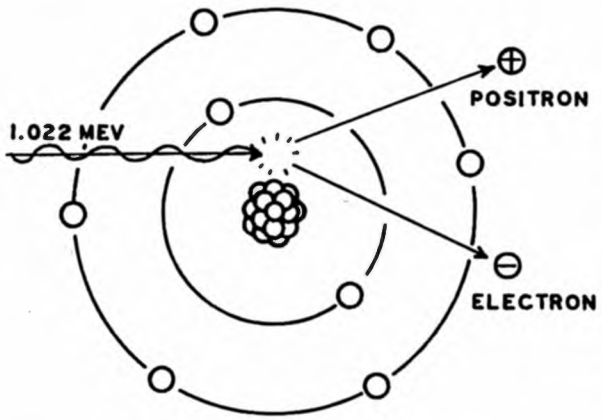


Fig. 29 Pair Production

D. Pair Production

When the energy of the incident photon is greater than 1.022 MeV, the photon may be absorbed through the mechanism of pair production. Pair production may occur when the photon closely approaches the nucleus of an atom and is subjected to the strong electrostatic fields of the nucleus. As the energy of the photon increases, the chances for pair production likewise increase. Although pair production occurs very close to the nucleus, no direct collision seems to occur.

As a photon approaches the strong nuclear fields, it ceases to exist as a photon; and a positive and negative electron pair appear in place of the original photon. These are known as a positron and an electron. This is a classical example of the conversion of energy (the photon) into two particles with mass (the positron and electron). This is governed completely by the relation due to Einstein: $E = mC^2$, discribing the direct conversion of energy into mass. In this relation, m is the mass created and C is the volocity of light.

It should be noted that the energy-equivalent of the rest mass of one electron is equal to .511 MeV. Therefore, for the production of 2 electrons (positive and negative), The reaction requires 1.022 MeV of energy. For this reason the photon must have this amount of minimum energy for the reaction to take place.

A nucleus must be present to provide the interacting electrostatic field for the reaction to occur. The nucleus does not enter into the reaction and no energy is left in the atomic system after the electrons have departed. This process actually must be considered as a theoritical collision between the photon and the nucleus and in this collision there is no measurable energy transfer to the nucleus.

If the energy of the incident photon is greater than 1.022 MeV, then the remaining energy in excess of that required to produce the electron pair is shared in various proportions between the two electrons as kinetic energy or energy of motion as they escape the system.

Kilovoltage	Photoelectric	Compton	Pair Production
10 KeV	95	5	0
20 KeV	50	50	0
40 KeV	20	80	0
100 KeV	1	99	0
200 KeV	0	100	0
2 MeV	0	99	1
24 MeV	0	50	50
100 MeV	0	16	84

Fig. 30 Relative Occurrence Rate in Water. Adapted from Johns, H.E., "The Physics of Radiology," 2nd Ed. 1961. Courtesy of Charles C. Thomas, Publisher, Springfield, Illinois.

E. Triplet Production

Triplet production is essentially the same process as pair production, except that the interaction occurs in the field of an electron. (Fig. 31)

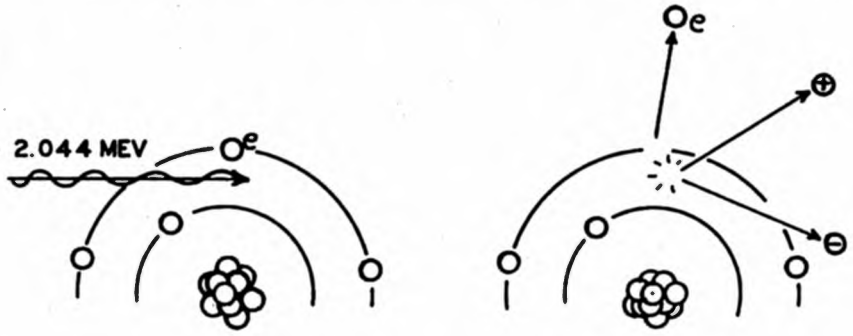


Fig. 31 Triplet Production

Thus, three particles appear in the interaction. The first two particles being the positron and electron of pair production, the third particle being the

involved electron of the atom. This electron is involved to the extent that there is recoil because of the identical mass of the three particles involved. A second principal variation between pair and triplet production is that triplet production requires a minimum energy of 2.044 MeV for its occurrence. Triplet production is a relatively rare occurrence and under the circumstances of high energy photons may account for a maximum of less than 10% of the positron production.

F. Fate of particles involved in pair and triplet production

Fate of the Electron: The electron which is created in the interaction of pair production will undergo the usual deceleration reactions of an energetic electron. (See Fig. 32) The length of time which it survives as an energetic particle will depend upon the amount of kinetic energy which was supplied to the electron at the time of its production. The electron will interact with other atoms until it has lost all of its kinetic energy at which point it will become a free electron at rest.

Fate of the Positron: The positron's ultimate fate is quite different from the electron. The positron as a free positive charge is unable to exist as such and eventually will undergo a process known as positron annihilation. For a more complete discription (See Section I-B-6).

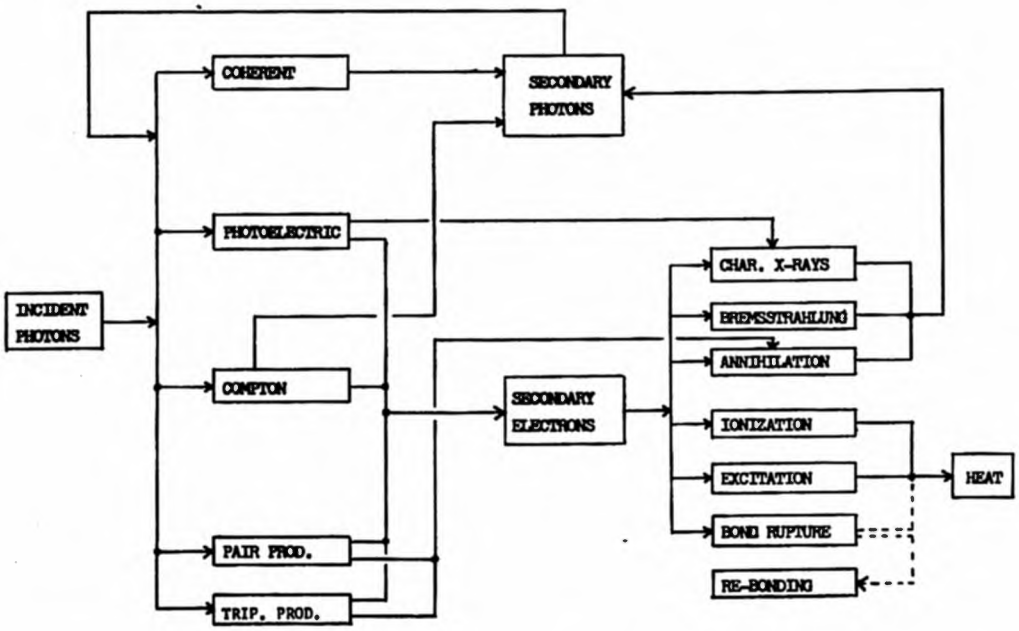


Fig. 32 Flow diagram of the Interaction and Eventual degradation products of an incident photon interacting with tissue.

VII CONCEPT OF QUALITY

A. Spectrum

In therapeutic x-ray the point of chief interest is the ability of the x-ray beam to penetrate and deliver its ionizing effects within a certain depth of tissue. The tissue penetration ability of a specific beam of x-ray is described as the quality of that x-ray beam or its degree of "hardness".

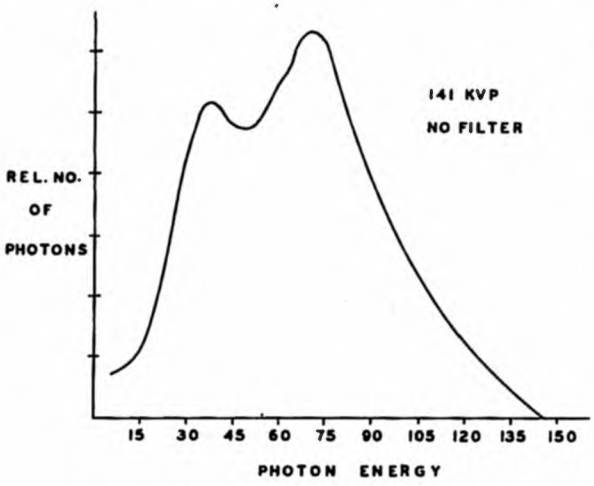


Fig. 33 Typical X-ray Spectrum

The quality of the homogeneous beam of radiation such as arises from gamma emitting sources is exactly and concisely known for any specific isotope and does not vary. With x-ray generators the quality of the beam is more complex, as the output of the typical x-ray beam is not monochromatic but rather is composed of a spectrum, or continuous distribution (Fig. 33).

This spectrum is made up of both Bremsstrahlung and characteristic x-ray. You will observe that a very small per cent of the spectrum is actually composed of the maximum kilovoltage energy, as very few of the electrons actually lose their entire energy in one single photon. Notice in Fig. 33, which is an experimental determination of a spectral distribution of 141 KVP x-rays, that a characteristic peak is reached at the "K" shell spectral energy of approximately 70 kilovolts. Below this "K" spectrum peak there is a second peak which is actually the continuous or Bremsstrahlung peak. It should be noted that if the peak kilovoltage had not been high enough for characteristic x-ray, we would have seen only the continuous peak.

B. Quality of X-ray

X-ray quality depends upon a number of factors, the following are the more important of these.

1. Voltage

The quality of the x-ray production depends to a great extent upon the voltage applied to the x-ray tube. The KVP applied to the tube will determine the peak acceleration of electrons and, hence, the peak voltage of the x-rays produced. Falling below this peak will be a spectrum as previously described, the general shape of this spectrum will be dependent to a large degree upon the wave form of the voltage as it is seen across the x-ray tube, (pulsating or constant potential).

2. Filtration

The second important factor which affects the quality of an x-ray beam is filtration. There is a certain amount of inherent filtration built into each x-ray unit over which we have little control. Beyond this, however, we are able to place external filters made of varying combinations and thicknesses of tin, copper, and aluminum. A particular combination of tin, copper and aluminum stacked in this order, from the tube to the patient, is known as a Thoraeus filter. The filter in an x-ray beam selectively removes a portion of softer x-ray and tends to harden the beam and thus increase the overall quality, however, at the same time reducing the quantity of output proportionally.

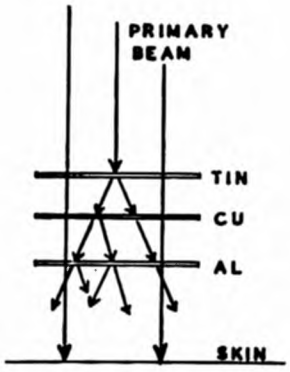


Fig. 34 Thoraeus Filter

C. Half-value Layer

In diagnostic x-ray we have an interest in the attenuation of the beam, In Therapeutic x-ray we have an interest in the absorption of the beam.

$I = I_0 e^{-ux}$

- I equals intensity of attenuated beam
- I_0 equals original intensity of beam
- e equals base of natural log (2.718)
- u equals rate of absorption of x-ray per unit thickness (expressed in cm -1)
- x equals thickness of absorber in centimeters

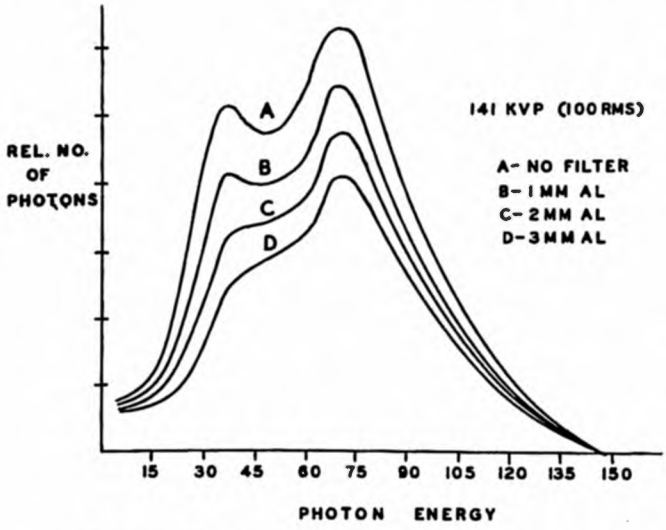


Fig. 35 The Effects of Various filters upon the x-ray Spectrum.

The linear absorption coefficient (u) may be defined as being numerically equal to the fractional reduction of beam intensity per centimeter of absorber. One can then demonstrate that the first one cm. will, for example, absorb 10% of the primary beam leaving an intensity of 90%. The next layer will absorb 10% of the 90% or 9%, leaving a total of 81% being transmitted by the two layers of absorber. It can be seen then as the layers of absorber are added, the reduction in intensity produced by each layer becomes smaller and smaller, as only a certain percentage of the remaining beam is removed each time. This is true in the strictest sense only for a monochromatic beam, as a degree of selective "hardening" is also apparent for a spectrum. This absorption scheme can be continued and it may be demonstrated from the decline that you can

theoretically never have an absorber sufficiently thick to stop all of the photons of the primary beam. In actual practice, however, a finite number of HVL may be used to reduce the beam to any desired value.

D. Measurement of Half-value Layer

The half-value layer is the thickness of an absorber required to attenuate half of the incident radiation, that is, to reduce the intensity of the beam to one-half its original value.

Although beam characteristics may be derived from a table, many of which are available, the exact characteristics of your particular machine cannot be determined from a table, as they depend not only on the stated values of the machine itself but upon the inherent filtration built into the tube head, the type and length of cables, and many other intangible characteristics that are particular to each and every x-ray machine.

A simple approach to this problem is to determine the half-value layer experimentally. From these readings one can then extrapolate and gain other characteristic information about one's own particular x-ray machine.

One should, of necessity, have a Victoreen chamber or other ionization chamber, in good operating condition and of reliable calibration for the quality to be measured. The chamber is placed at a very exact distance within a comparatively small treatment field, 5 x 5 cm or smaller. A part of the chamber body is usually included in the treatment beam and this characteristic should be taken into consideration with the position of the chamber duplicated after each reading. From previous calculations one will have a comparatively good idea of the half-value layer of the machine and a selection of copper absorbers covering this general range should be available.

A number of initial chamber readings are taken with only the machine's filters in place. These readings are then plotted on a 1 cycle semi-log graph as 100 % transmission. All filters and the treatment cone to be used must be in place as it would be invalid to attempt to determine half value layer without the set-up being exactly as it would be used.

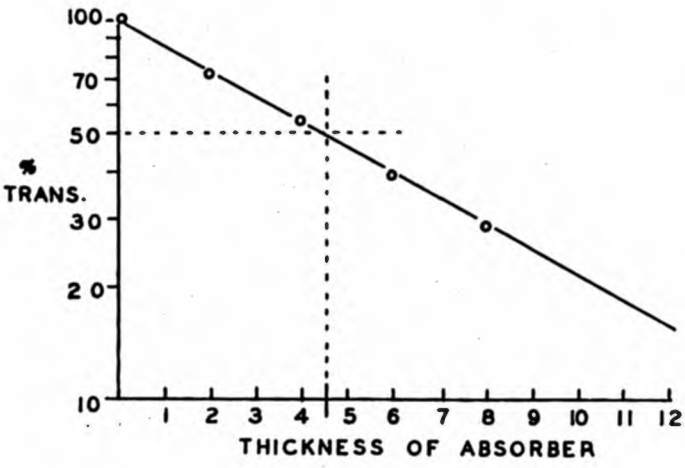


Fig. 36 Half-value Layer Determination

Following this, multiple determinations are carried out utilizing adequate exposure times and increasing the number of thicknesses of absorber. These values are then continued as a plotting on the graph and the half-value layer may then be determined when the points are connected with a line.

VIII CONCEPT OF QUANTITY

A. The Roentgen

In 1928 the roentgen based on ionization in air was adopted as a unit of x-ray quantity. The roentgen until recently was defined as the quantity of "X" or gamma radiation, such that the associated corpuscular emission per 0.001293 grams of air produces, in air, ions carrying 1 ESU of quantity of electricity of either sign. This mass of air is 1 cc at 0 degrees C. and 760 mm of mercury pressure. The roentgen has now been re-defined by the ICRU at its meeting in Montreux, Switzerland, April 1962, and published in the ICRU Handbook 84, the special unit of exposure is the roentgen (R).

$$1R = 2.58 \times 10^{-4} \text{ Coulombs/kilogram}$$

Roentgens, of necessity, must be measured in free-air ionization chambers. Although this is somewhat cumbersome, it can be done with exacting precision. However, very few people have free air ionization chambers, therefore, Victoreen and other air equivalent wall chambers have been developed over the years. Although these chambers do not measure with the exactness of the free air chambers, they are well within clinical limits and, in addition, may be calibrated against free air chambers for greater accuracy.

In order to satisfy the definition of the roentgen, however, it is necessary that measurements be carried out under conditions of equilibrium between the primary x-ray and the secondary electrons. Up to a few kilovolts this is the situation with standard ionization chambers and free air chambers. With air wall equivalent thimble chambers of suitable thickness this equilibrium is also satisfied. Above the million volt energy level new phenomena begin to take place which make it increasingly difficult to adhere to the conditions of measurements in roentgens either in free air or in air wall equivalent chambers. Secondary electrons become much more energetic and positron-electron pairs begin to appear with annihilation radiation producing energetic secondary photons. The roentgen finally becomes unable to be realized in this region and above. The roentgen was not applicable to any other type of radiation, such as alpha, beta or neutron particulate radiations until the new definition was agreed upon.

B. The Rad

The ionization produced in a small volume of air is proportional to the energy that would be absorbed by tissue exposed at the same point if the atomic number of the tissue was the same as that of air. We could then express our exposure in roentgens.

Unfortunately the effective atomic number of tissue is not equal to that of air and the pattern of secondaries is also very different. This difference becomes more and more significant as the energy ranges are increased. Changes can be caused only by the deposition of energy and this occurs via secondary electrons.

It was decided at the 7th International Congress of Radiology in 1953 to recommend a unit of absorbed dose to express energy deposition. The rad (Radiation Absorbed Dose) was the result.

The quantity of radiation delivered to a certain place is called the exposure and is measured in roentgens. The quantity of energy absorbed from the exposure is the absorbed dose. The unit of absorbed dose is the rad. One rad is 100 ERGS per gram of any absorber. In practice it is still necessary to measure the output of any x-ray unit in terms of the ionization produced in a thimble chamber with the scale graduated in roentgens. Calculations can then be made taking into account the quality of the radiation and the atomic number of the soft tissue, these factors being utilized in the calculation of a rad to roentgen ratio.

Where a dose is quoted in rads, what is usually meant is absorbed dose if the tissue at that point were soft tissue of average composition. In special cases where the area of interest consists of a large amount of bone or fat then this must all be taken into consideration in the calculations.

The energy absorption of a given medium depends upon a number of factors. Among these principally being the photon energy and the atomic number of the absorber. As noted in Figures 37 and 38 the absorption in the low KV range is markedly higher in the bone than in low atomic number fat, whereas the absorption in muscle remains relatively constant throughout the entire energy spectrum as compared to bone or fat.

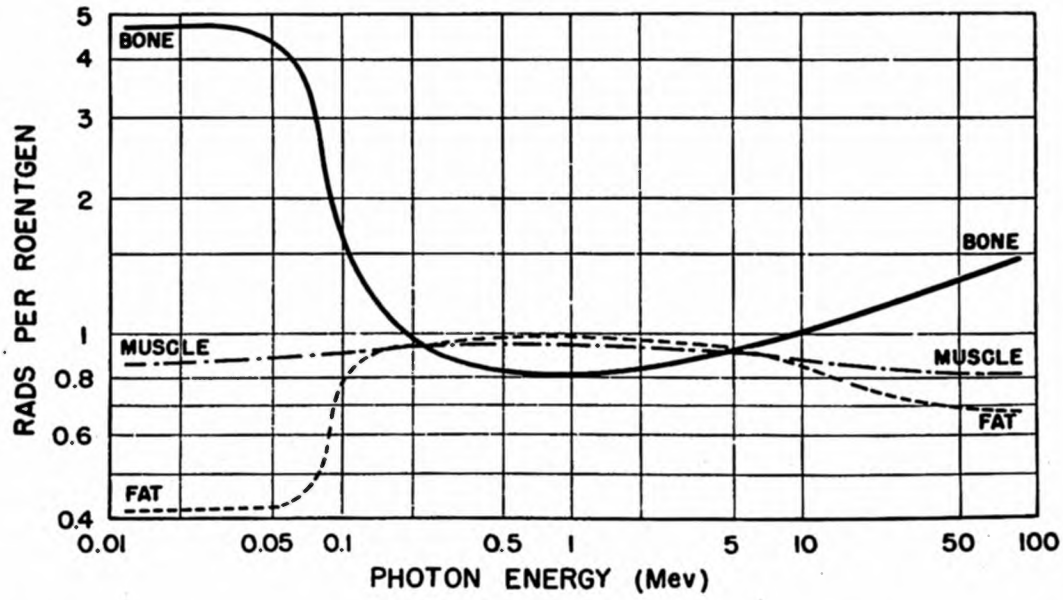


Fig. 37 Relative absorption in bone, fat, and muscle as related to energy of the incident photons. Reproduced from Glasser - "Physical Foundations of Radiology" 3rd Ed. Paul B. Hoeber Publisher, New York, New York.

Tube potential	*Filter*	hvl	f = rad/roentgen (*Average values*)			
kv.	mm. Cu	mm. Cu	*Water*	*Muscle*	*Bone*	*Fat*
100	0.18	0.25	0.91	0.94	3.10	0.64
200	0.2	0.5	0.92	0.94	2.69	0.72
250	1.0	2.0	0.96	0.96	1.42	0.87
400		4.0	0.97	0.97	1.11	0.97
Co-60			0.97	0.98	0.87	1.01
Betatron (2.5 Mev.)			0.96	0.96	0.95	0.95

Fig. 38 Average absorbed dose per roentgen of exposure dose in water, muscle, bone, and fat for various primary x-ray spectra. Reproduced from Glasser - "Physical Foundations of Radiology" 3rd Ed. Paul B. Hoeber Publisher, New York, New York.

IX CALIBRATION

The calibration of your x-ray machine is one of the most important maneuvers which you will carry out prior to treating the patient. This calibration must be done with great accuracy as any error introduced at this point will simply be compounded manifold as the patient is treated each day. It is advisable, unless you are an expert in this field, that your machine be calibrated at least once a year, and after any major x-ray circuit work, by a qualified radiation physicist. These readings, however, should not be relied upon blindly for the remaining period of time that the machine is used, as either subtle or acute changes may occur within the x-ray circuit which could cause dosage error. For this reason you should spot-check your therapy machines at a standard setting at least once a week to detect any gross calibration difference. The therapy machine should have an accurate calibration carried out by you or a member of your staff at least once every two months utilizing an average set of factors commonly used in the carrying out of your radiation therapy.

A. Method

The exact steps for calibrating an x-ray generator are very completely outlined in Johns "The Physics of Radiology", however, the following points should be carefully observed:

1. Select either a specific cone of interest or a convenient field size, usually a 5 x 5 cm field. This field is large enough so that slight placement errors of the chamber will be relatively unaffected and not so large that scatter problems become significant.

2. The therapy beam should be preferably directed horizontally so as to have a large air volume beyond the cone to eliminate backscatter from hard objects such as the table top or the floor.

3. The Victoreen chamber should be mounted in a clamp so that the sensitive volume of the chamber is in the exact center of the treatment field with its axis perpendicular to the beam axis and the center of the chamber at exactly the selected focal distance. Select the desired machine characteristics for which calibration is to be determined. If the approximate half-value layer for these particular settings is not known, it may be determined experimentally as previously outlined.

4. Select a Victoreen condenser chamber which is designed to measure this quality of radiation.

5. Adjust the chamber for "zero" reading in the Electrometer, place the chamber, in its cap, in the holder within the center of the treatment field. Always keep the chamber capped and never allow anything to touch the contact area. Even the cleanest hands may leave a conductive residue on the insulating surfaces and cause errors of calibration.

6. Turn on the machine and bring it to the operating conditions previously selected, set the timer to an appropriate value, and open the shutter. The exposure time should be sufficient to discharge the chamber approximately 60%. Several exposures may be necessary to determine the correct time. All machine factors should be kept constant during the calibration period.

7. Remove the chamber from the cap and insert it into the electrometer. Adjust the index of the electrometer as per the instructions, insert the chamber all the way into the electrometer and record the reading.

B. Shutter Error

When calibrating an x-ray machine so that it will be accurate over the entire range of time and output one must, of necessity, correct for shutter error. This is the amount of error that is inherent in the act of opening and closing the shutters on the machine, if it is so equipped. The same error is present on isotopic teletherapy type machines. On many Cobalt units and some x-ray machines this may be 2 to 3 seconds. This shutter opening time may be unimportant, however, this must be determined experimentally before any decision can be made. It should be remembered that a 3 second shutter time is 10% of a 30 second exposure, however, it is only a negligible fraction of a 10-minute exposure.

Many of the smaller x-ray units do not have a shutter and, therefore, the only error which would need correction is the human error of turning the machine on and off.

The simplest way to correct for shutter error or to determine the exact time elapsed during the opening and closing of the shutter is to take three specifically timed readings.

For example:

17 seconds exposure equals 13.00 R on the Victoreen scale.
20 seconds exposure equals 15.55 R on the Victoreen scale.
3 seconds exposure equals 2.45 R on the Victoreen scale.

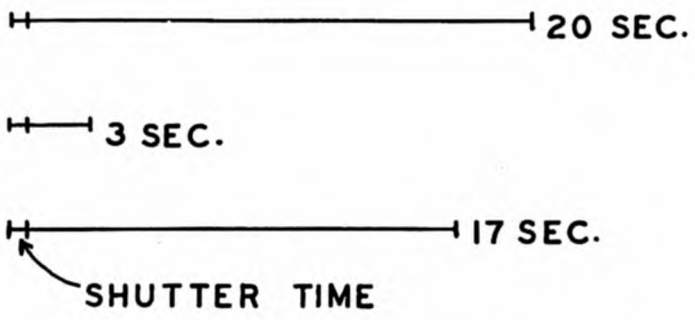


Fig. 39 Relationship of shutter time to "on" time

Subtract the readings of the 3 second time (2.45) from the 20 second reading (15.55), this cancels out the shutter error, and gives a true 17 second reading of 13.10.

Compare this reading (13.10) to the actual reading at 17 seconds (13.00) .

$$\frac{13.10R}{17s} = .77R/sec; \quad \frac{.1R}{.77R.sec} = .13 \text{ second shutter time}$$

We, therefore, have arrived at the correction for shutter time, namely, -.13 seconds. Subtract, in this case, the correction from time indicated to achieve true "on" time.

C. Compensation for Temperature and Pressure

With sealed chambers or with some of the new chemical or solid state dosimeters corrections for temperature and pressure are not necessary. However, the Victoreen chamber is an unsealed air chamber and is adjusted to read correctly only at 22° C and 760 mm of mercury pressure. For this reason temperature and pressure corrections must be carried out as outlined in your Instruments Calibration Manual. It should be noted that the barometric pressure required is the station pressure at the location of the Victoreen, not that corrected for sea level as is commonly reported by the local weather bureau.

D. Calculations

Now that we have arrived at the output of the machine minus the shutter operating time, we can set about using the following example to calculate the

exact output in roentgens per minute from the machine, utilizing all other corrections for temperature and pressure and inherent instrument calibration factors if known.

$$13.1R \times \left(\frac{60s}{17s}\right) \times 1.029 \times \left(\frac{273 + 22.9}{273 + 22.0}\right) \times \left(\frac{29.92}{28.60}\right) = 50.0\ R/m$$

Bureau of Standards correction = 1.029 for our instrument
Temperature correction (room temperature = 22.9C)
Pressure correction (local pressure 28.60 in. Hg)
29.92 = 760 mm Hg

F. Cobalt 60 and Cesium 137 Calibrations

Most of the items discussed in the calibration of an x-ray machine apply to an Isotope machine also. Again, one must establish the factors for the particular field being calibrated. With an isotopic machine, however, one does not have the problems of filters, KV, and MA to observe. The significant factors which may vary on an isotope machine are field size, SSD, time, shutter time, and decay. When measuring Cobalt 60 or Cesium 137, the wall of the ionization chamber must be thick enough for equilibrium conditions to exist. Select a particular chamber from your selection which is specific for use in the energy range which you are measuring. Most ionization chamber units have either a chamber with a very thick wall or sleeve to fit over a specific chamber when utilizing higher energy radiation in the region of Cobalt 60.

When calibrating a betatron or other very high energy machine one must, of necessity, go to more complex calibration methods utilizing a very thick equilibrium wall so that the Bragg-Gray principle may be utilized. The cavity in which the chamber lies should be located at the area of the build-up peak.

X FACTORS INFLUENCING DEPTH DOSE AND BACKSCATTER FACTOR

The determination of the depth dose or the dosage at a specific measured distance within the object being irradiated is one of the most important parts of x-ray therapy dosimetry. An important fact to be remembered is that it is the actual dosage within the depth of the tumor which we are treating that is of the greatest importance in x-ray therapy. Determination of tissue or depth dose involves determination of the depth of the lesion below the surface of the skin for each portal being treated.

When tumor depth is known many charts and tables are available which give the number of rads or roentgens delivered to any specific depth, usually expressed in the percentage of the dose either in air, on the surface, or at a point of maximum build-up. These tables are referred to as depth dose charts or tables.

An x-ray machine is ordinarily calibrated so that its output is known in roentgens per minute at a specified position. If the body of a patient, or a phantom, is introduced into the beam so that the skin or the surface of the phantom is at this same position, the number of roentgens per minute is increased

owing to radiation scattered back from the underlying material. For 250 KV x-rays and moderately deep tumors, the depth-dose may well be about 10% primary and about 90% secondary (scattered radiation).

The significance of backscatter or the backscatter factor is in calculation of a maximum dose. This generally will be the dose in roentgens or rads at the surface or near the surface, depending upon the quality of the beam, and is always calculated at the center of the field. In the case of low energy x-ray machines, the skin exposure dose will be the maximum dose and may be measured with a suitably calibrated thimble or extrapolation chamber. The maximum exposure dose may also be calculated from the exposure dose tables if the backscatter factor is known. Under these circumstances it may be calculated for any half-value layer from extremely low KV up into the super voltage range. The maximum exposure dose is the sum of primary beam and the backscatter. It varies markedly with the quality of the radiation and with the volume of backscattering material present. As a general rule the harder the radiation beam the greater the tendency for the scatter to be in a forward direction, and conversely, the softer or lower the energy the more likely the scatter will be either in a side or back direction. Below HVL 1/2 mm Cu, the per cent backscatter decreases because photoelectric effect is becoming vastly more important than Compton effect.

There are multiple factors which influence depth dose. For a beam of x-ray the percentage of depth dose depends chiefly upon:

A. Radiation Quality
B. Depth below the skin surface
C. Area of the field
D. Thickness of tissue and support directly beneath the part irradiated
E. Target skin distance

A. Radiation Quality

The scatter photon production becomes more forward as the energy value increases. With soft x-ray at approximately .5 mm copper half-value layer, between 40% and 50% of the scatter will be backscatter. Fig. 40. As the energy of the radiation increases the degree of backscatter becomes less and the scatter becomes more forward until the energy of Cobalt 60 is reached where most of the energy is in a forward direction and the backscatter factor will be about 3%.

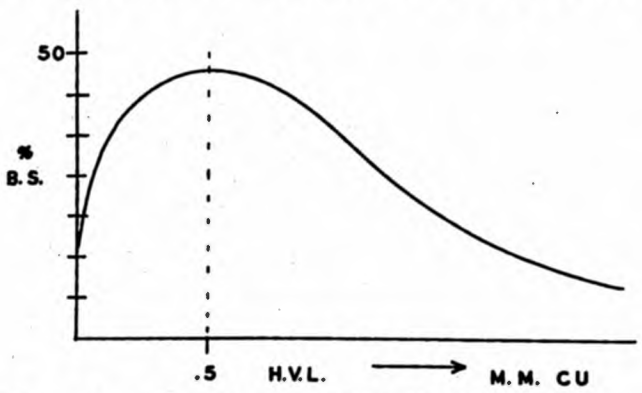


Fig. 40 Backscatter as a function of HVL

There is no simple method for calculating the effects on depth dose by changing KV or filtration. Such data must be obtained experimentally with each individual machine. Increasing either kilovoltage or filtration will increase the penetrating power of the radiation hence it will increase the relative depth dose.

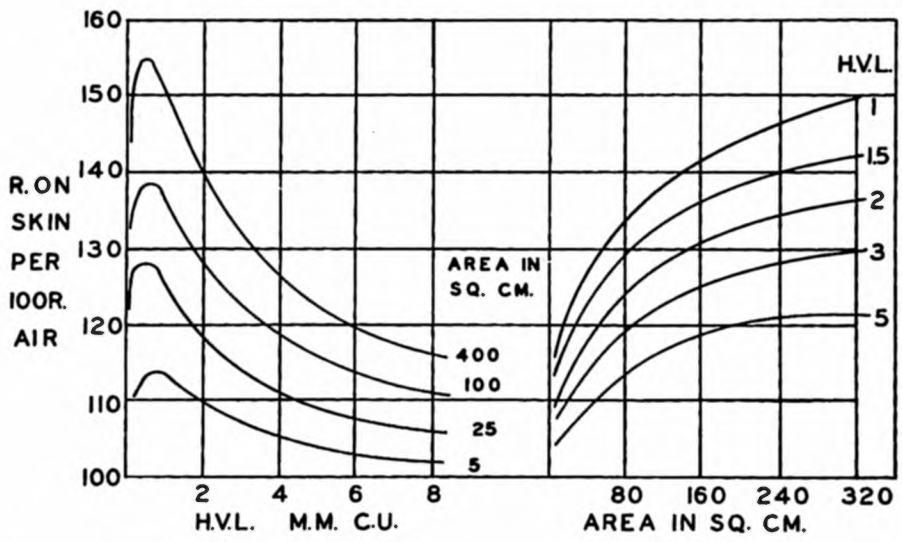


Fig. 41 These curves show the variation of surface dose (air dose plus backscatter) with variations in irradiated area and quality of radiation. Adapted from Glasser, O., "Physical Foundations of Radiology," 3rd Ed. 1961, Courtesy of Paul Hoeber, Publisher, New York, New York.

B. Tumor Depth

One will find that the deeper the lesion the less the depth dose for a given surface exposure. This logically is due to both absorption and scattering in the overlying tissues as well as increased target tumor distance. All of these factors, therefore, work together to create a lessening of the primary beam of radiation as material is traversed. (Fig. 42)

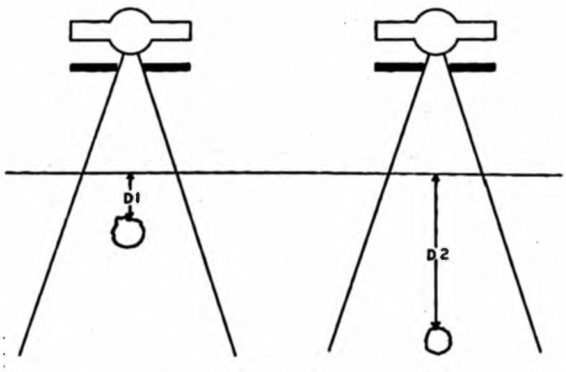


Fig. 42 Variation in depth dose due to depth of tumor

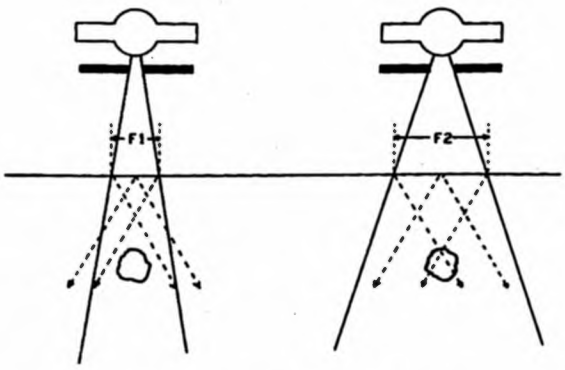


Fig. 43 Variations in depth dose with field size

C. Size of Treatment Field

It may be shown (Fig. 43) that the larger the area of the treatment field the greater will be the scattering up to a maximum. The maximum field size depends primarily upon the range of secondary or scattered radiation which depends in turn upon the energy of the primary beam. At any given depth, and with a specific quality of the primary beam, the depth dose will be greater for a larger field than for a smaller one. This apparent increase in depth dose is due to the fact that both of these fields will receive the same amount of primary beam radiation per unit area however, with the larger field, there is a greater chance for scattering toward the tumor and, hence, the depth dose at this particular point will be enhanced by this scattering effect. This effect becomes more noticeable as the area of the field increases. The backscatter effect reaches a peak at approximately 200 kilovolts and is less important on both sides.

For every quality of radiation the backscatter increases with an increase in the area being irradiated with the effect being less for the higher energy x-ray than for the soft.

D. Variation with Backscattering Volume

Thickness of tissue and support directly beneath the part being irradiated is very important. At or near exit portals the absence of backscatter reduces the dose markedly below values found in depth dose tables which are usually derived from thick phantoms. (Fig. 44)

If one is given a field of a definite size with all other factors being held constant, one will find that the larger the volume or the more dense the underlying tissue, the more secondary or backscatter there will be to affect the depth dose at the level of the tumor. This can easily be seen as the underlying tissue will create considerable backscatter and logically the larger the volume of underlying tissue the more backscatter may be created. Underlying bone will give a much higher backscatter because of its greater density than underlying soft tissue. This rule holds true in the lower or ortho-voltage ranges, but becomes less important as one reaches the super voltage ranges (Cobalt 60 and greater). With the

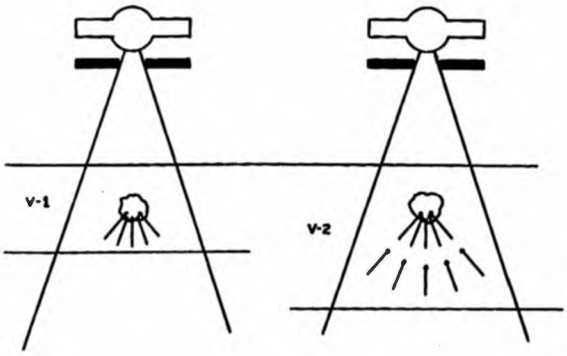


Fig. 44 Variation in Depth Dose with Scattering Volume

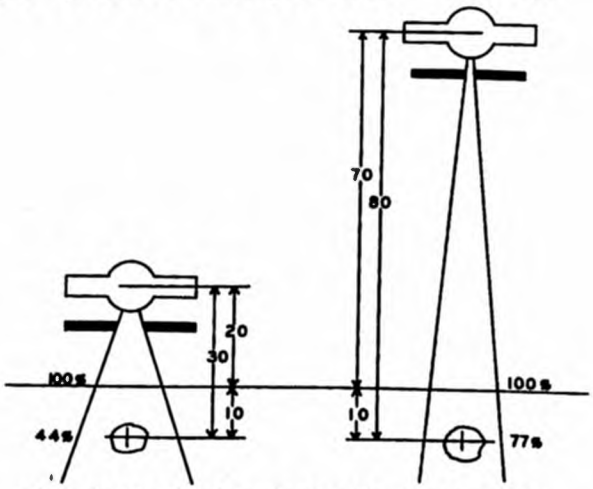


Fig. 45 Variation in Depth Dose with increased target skin distance

higher energy ranges the majority of the scatter is forward and, hence, there is less effect from this specific mode of backscattering.

E. Target Skin Distance

The difference in intensity for 2 points separated by 10 cm. (skin to tumor distance) is considerably greater if all of the distances are relatively short, than if all of the distances are relatively long. For example: (Fig. 45) from 20 to 30 cm. the intensity decreased 56% by inverse square law, however, from 70 to 80 cm the intensity decreases only 23% by inverse square law. This comparison holds quite true when considering distances either in air or within an absorbing and scattering media such as the body, as far as the primary beam of radiation is concerned. Absorption of radiation by the 10 cm. layer of tissue is the same in both cases. The only true difference then is the geometrical relationship between the source, the skin, and the lesion. This effect becomes more important as the energy is increased, due to the fact that the majority of the dosage reaching the tumor is primary beam and forward scatter with less dependence upon side or backscatter. As the energy increases the per cent of effect on depth dose due to the primary beam becomes greater and the effects of scattering become much less.

XI ISODOSE CHARTS

A. Inverse Square Law

The intensity of a beam of photons is inversely proportional to the square of the distance. Divergence of the beam as it traverses the tissue is dependent upon the inverse square. As in Figure 46-A one will see that a given population of photons will be distributed over an area of a field with a certain photon density. This area of the field may be thought of strictly in terms of a particular segment of the surface of a sphere. Each time the radius of the sphere is doubled, the actual size of the area which must be covered by the specific photon density will increase 4 times. This is the law of the isotrophic emission of finite objects.

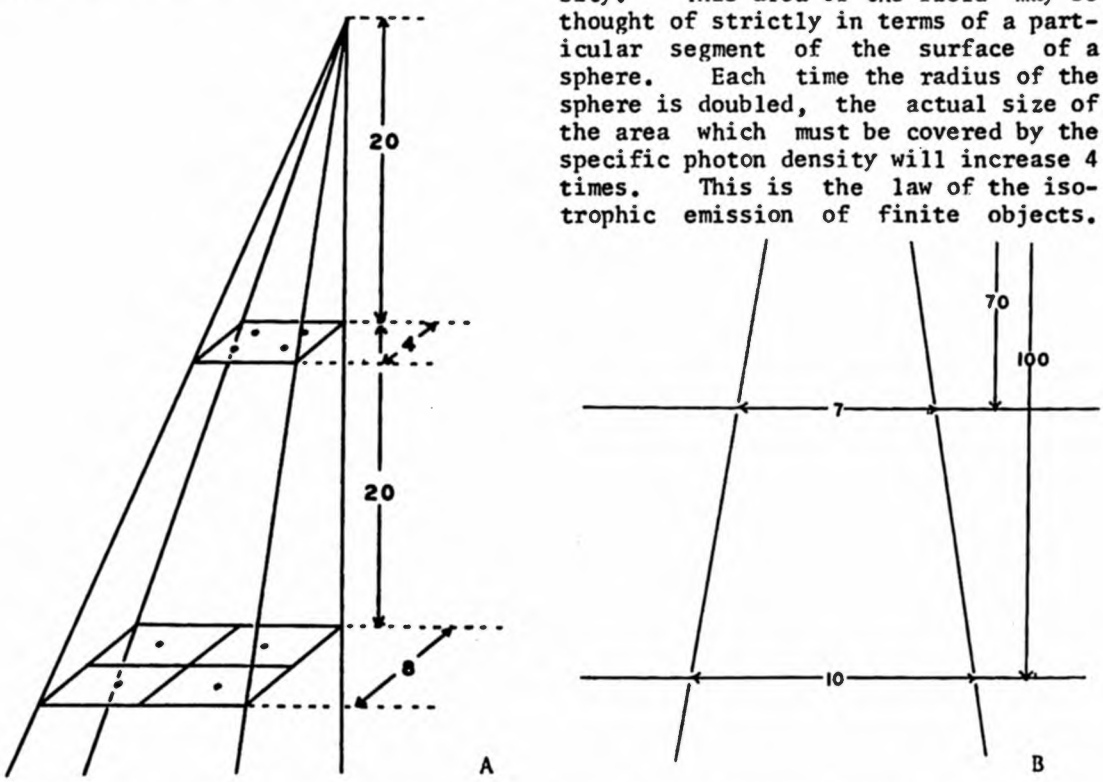


Fig. 46 The Effects of Distance Upon Dose Rate - Inverse Square Law

$$\frac{I_1}{I_2} = \frac{D_2^2}{D_1^2}$$

This law works for radiation, non-focused light, explosions and other expanding spheres of a similar nature. A practical example with which we must deal daily is seen in Figure 46-B. A given field 7 cm square at 70 cm target skin distance will diverge to a field 10 cm square at a depth of 100 cm target skin distance. The isodose curve that results from the interaction of this particular beam with the tissue is dependent to some degree on the inverse square drop-off effect and this must be taken into account.

B. Penumbra Effects

Penumbra accounts to a great extent for the rounding that is present on the lateral edges of isodose curves. Backscatter plays an important part in this but penumbra accounts for the majority of the effect. The beam from a Cobalt 60 unit is almost all primary with about 10% or so of the beam being scatter within the source housing and field defining system. As can be seen in figure 47-A considerable softness or fuzziness exists along the edge of most fields treated with an isotopic teletherapy machine due to the relatively large size of the source. This effect can be minimized by placing the diaphragm much closer to the skin. This becomes quite cumbersome, however, and in many cases very impractical. For this reason a number of manufacturers have devised a secondary diaphragm which is retractable. This is called the penumbra trimmer and usually consists of four bars of uranium, tungsten alloy, or lead, which may be extended quite close to the patient as in figure 47-B. These trimmers succeed in shaving considerable of the penumbra from the field and results in a much sharper definition of the field.

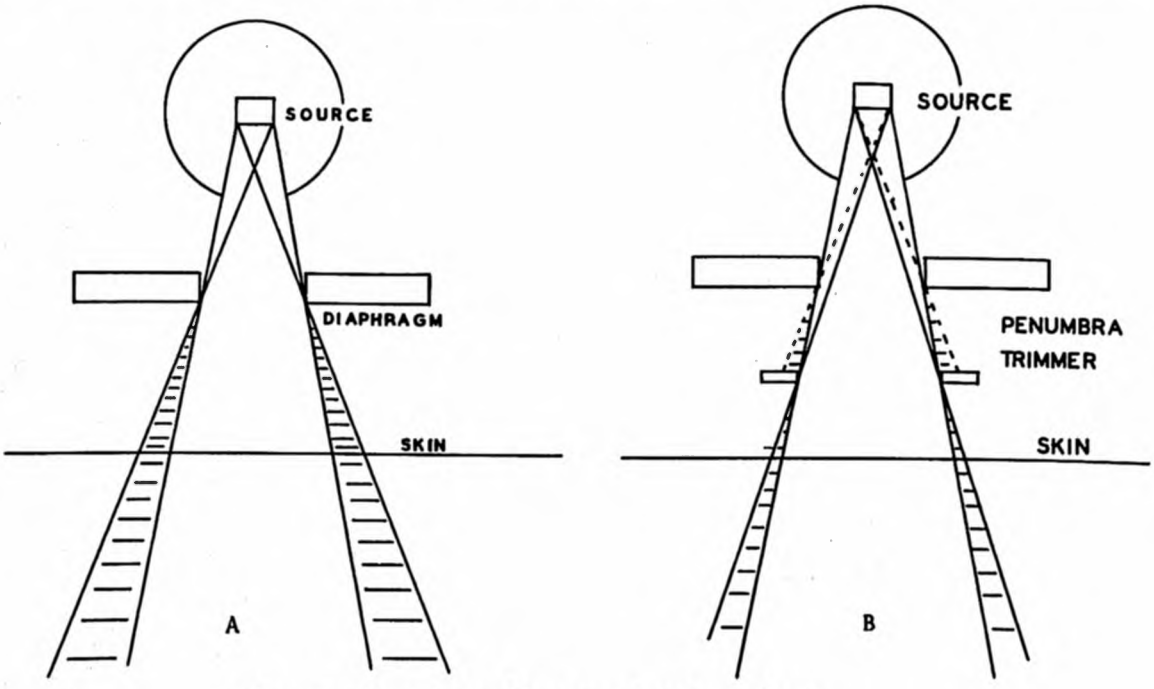


Fig. 47 Penumbra effects and trimmer bars

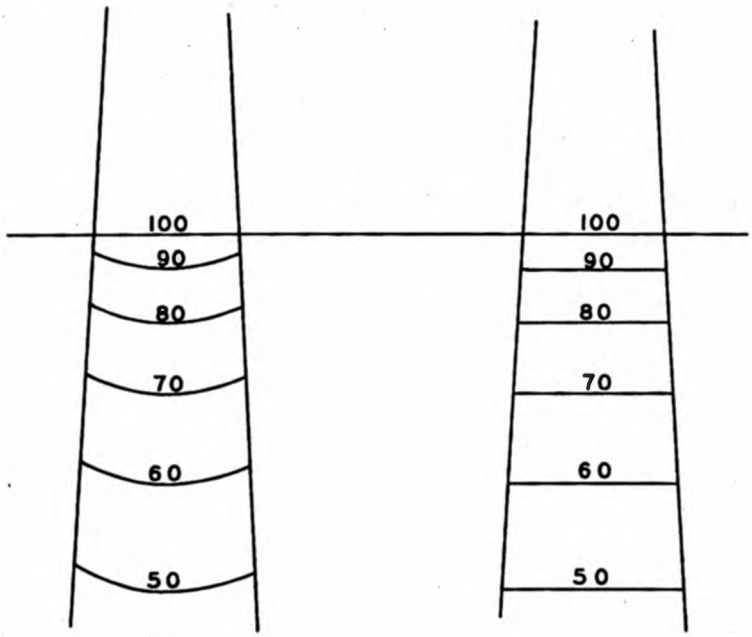


Fig. 48 Isodose Curves

C. Isodose Curves

Isodose curves are constructed of isodose lines, which are points of the same dose rate connected with a single line. These usually are arbitrarily set at either 5 or 10 per cent intervals, however, any acceptable interval may be used. Isodose lines characteristically have a rounded central appearance with an upsweep to the edges. (Fig. 48-A) The edge upsweep is due almost entirely to the loss of side scatter radiation at the edge and the fall off due to penumbra effect. Isodose curves would be practically square if there were no scatter or penumbra effects noted. (Fig. 48-B) This is the reason that the isodose curves do become more nearly square as the energy is increased. There would be only a very slight downward bow in the center of the curve due strictly to inverse square if this were the only effect that were operable. This difference would amount to approximately 1/2 of 1% difference at the edge of the field as compared to the center. Most isodose curves are derived from automatic isodose plotting in a water or wax phantom which is approximately tissue equivalent. Film has been used in a number of instances for plotting of isodose curves, however, film is markedly dependent upon the energy of the beam and the energy of the secondary scatter, all of which must be taken into very careful consideration when attempting to plot curves in this manner.

Isodose curve manipulation may be carried out with a fairly good degree of success when utilizing higher energy x-ray in the range of 1 MeV or greater. Isodose manipulation is particularly successful when utilizing the homogenous beam of Cobalt 60. It is very difficult to use wedge filters with very low

energy KV, because of the tremendous amount of back and side scatter that is created within the filter. With Cobalt 60, however, it is possible to attenuate the beam by the use of in beam filters and arrive with very accurate per cent reductions of various parts of the beam. The utilization of wedge filters has found great application in treatment programs where multiple ports must be used to treat an area while avoiding nearby vital structures. The generation of isodose wedges is beyond the scope of this particular session and should be attempted only by those who are familiar with the procedure.

D. Electronic Equilibrium

Electronic equilibrium may be defined as the point within an absorbing and scattering media where maximum ionization from secondary electrons plus the primary beam occurs. Prior to the point of electronic equilibrium, the same number of electrons are being formed as are being absorbed, beyond the point of electron equilibrium more electrons are being absorbed than are being formed. The region between the surface and the point of maximum dose is called the build-up region. The portion beyond this area is the region of electronic equilibrium. The depth for the electron build-up maximum is the average range of the electron in the material being considered.

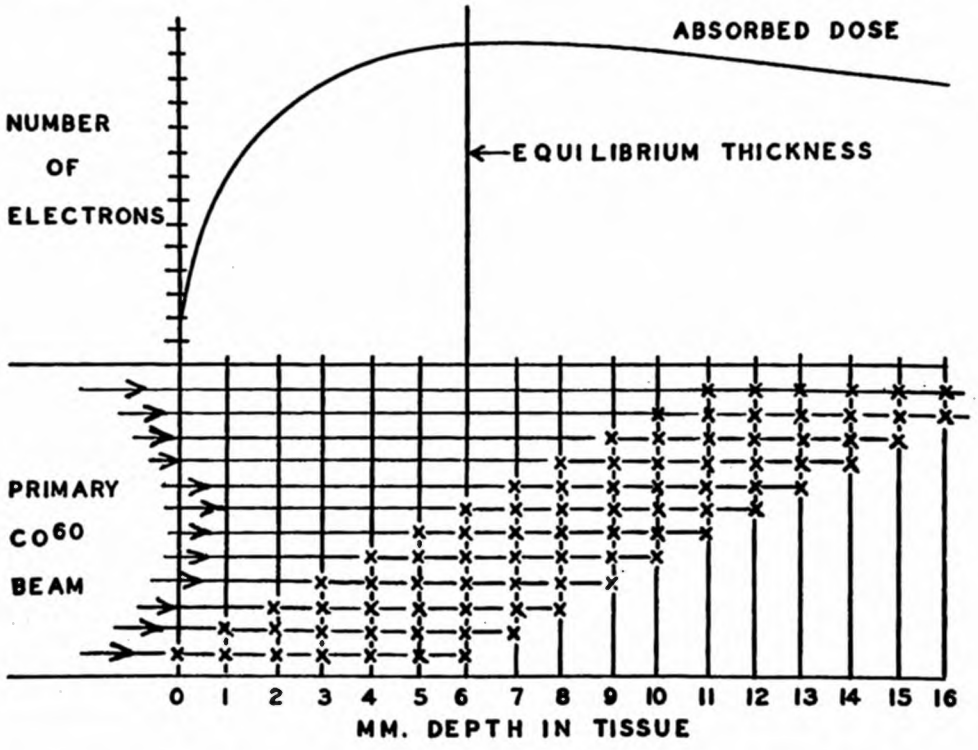


Fig. 49 Electronic Equilibrium. Adapted from Johns, H.E., "The Physics of Radiology," 2nd Ed. 1961. Courtesy of Charles C. Thomas, Publisher, Springfield, Illinois.

For true electronic equilibrium to exist one must consider that the primary beam is not attenuated in the area of interest. This, in fact, is seldom true. An equilibrium thickness may exist which is a fine region of electronic equilibrium beyond which there is drop-off. Beyond the equilibrium thickness, the absorbed dose, will fall approximately exponentially. For a true electronic equilibrium to exist the per cent attenuation of the primary beam through the distance of the range of secondary electron must be extremely small. This is the case up to approximately 500 KV, however, as the energy continues to go higher the attenuation of the primary beam becomes significant as related to the track of the secondary electron until a point is reached at approximately 3 MeV where the secondary electron track is far longer than the area over which the primary beam may be considered unattenuated. This is the principle reason for recommendations that the roentgen not be used as a unit of exposure for energies beyond 3 MeV, as the range of secondary electrons in air is so great that a practical chamber to measure the roentgen as defined is impractical.

E. Central Axis Depth Dose Curves

This type of curve will be derived if one plots the depth on the central axis of the beam as related to per cent of some normalized dose remaining at these particular depth plots. This will appear as a curve which for Cobalt 60 will have approximately the appearance of Illustration 50. The 100% line always being reached somewhat below the surface due to electron buildup.

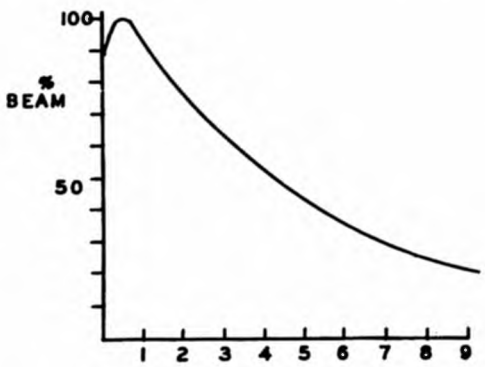


Fig. 50 Central Axis Depth Dose

If, however, central axis depth dose is plotted on semi-log paper (Fig. 51) with a ratio of centimeters in depth along the central axis as related to the log of the per cent of the dose, we find we have almost a straight line once the maximum dosage has been reached below the surface.

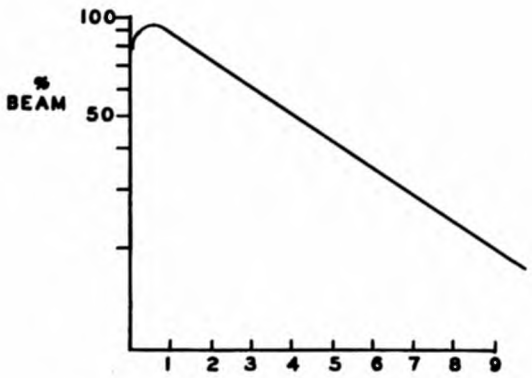


Fig. 51 Central Axis Depth Dose (Semi-log Plot)

If one stays near the center of the beam you will see very little effect by either scatter or penumbra. For that reason this line may be considered almost a perfectly straight line. The reason this is a log function is that fractional reduction of intensity is proportional to the intensity. It should also be noted that the fractional reduction of intensity is a constant, therefore, the absolute reduction in intensity depends upon the quantity that is present to be reduced.

XII SUPER-VOLTAGE THERAPY MACHINES

A. High Voltage Conventional X-ray Machines

Conventional x-ray machines consist, basically, of a high voltage stepup transformer with an x-ray tube across it. The x-ray tube is usually of a relatively simple design consisting of a hot filament as a source of electrons and an anode or target for the generation of x-rays. The acceleration of the electrons across this tube is accomplished in one step without any grids, guides, or other mechanisms other than the electron focusing design of the cathode. The electrons accelerated across the gap between the cathode and the anode achieving their maximum energy at the time of collision with the target. These machines are usually effective up to about 440 kv. Some units have been built as high as 660 kv using conventional tube principles. The majority of radiation therapy x-ray machines up until about 1945 were of this particular design. A few 1 million volt machines had been built just prior to 1940 utilizing an entirely new tube design known as the accelerator tube. The majority of conventional x-ray units frequently termed "orthovoltage", generally use a grounded center tap secondary to the high voltage transformer which leaves both the anode and filament of the tube at a high potential above common ground, usually one-half of the effective kilovoltage of the tube. The tube is immersed in a coolant oil bath and is well insulated from surrounding structures.

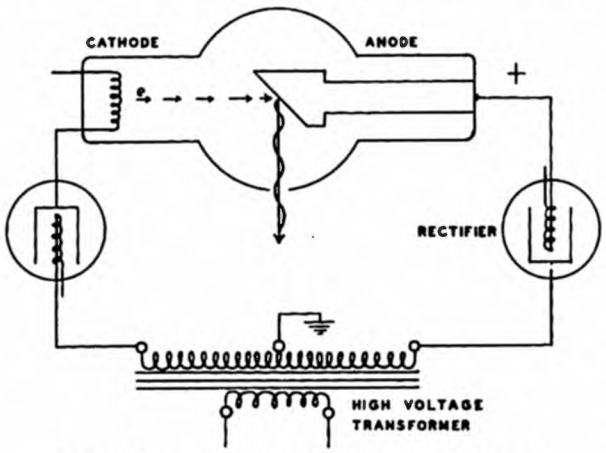


Fig. 52 Conventional X-ray Machine

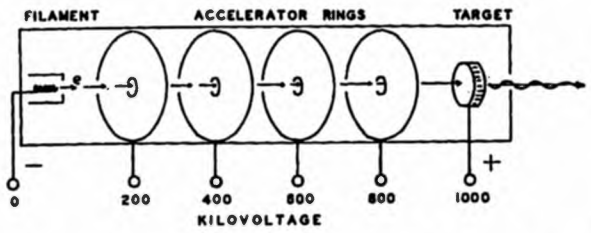


Fig. 53 Accelerator type X-ray Tube

B. Accelerator type X-ray Tubes

Because of the electrical stress limitations placed on conventional x-ray tubes by very high voltage, these tubes have a tendency to arc over across the glass or through the insulating barriers.

Due to the ability of very high voltage to arc through the vacuum inside a conventional x-ray tube, the distance inside the tube must be increased to get adequate separation between the filament and the anode to the point where reliable and reproducible electron acceleration cannot be achieved by simple means.

The electron accelerator type x-ray tube was designed specifically for use with high voltage transformer type machines of above 1000 KvP. This tube consists of filament, a series of accelerator rings, and a target at the far end of the tube. (See Fig. 53). In this type of tube the accelerator rings are connected to

equally increasing steps of positive high voltage until the final ring is only slightly less positive than the anode. The potential difference between any two accelerator portions of the tube in figure 53 is about 200 kilovolts, which is less than the breakdown limits of the tube. In this example, electrons created at the filament will be accelerated toward the first accelerator ring with an attraction force of 200 kilovolts. Upon reaching the region of the ring the electrons have gained considerable momentum and readily pass through the circular opening in the center of the accelerator. Once they have passed this area they are now attracted by the second ring which is 200 kilovolts more positive than the one they just passed through. This process is repeated through each step down the tube with the electrons gaining an energy of 200 kilovolts with each stage. At the final ring they pass on toward the anode eventually striking it with the full cumulative energy of 1 MeV.

Most tubes of this type have what is referred to as a transmission type target; the x-rays produced are transmitted through the target to become the useful field. A transmission target is used due to the angular distribution characteristics of x-rays created at these high energies. Polar diagrams have been calculated showing the variation of the intensity of x-ray with angular distribution as a dependency upon electron acceleration kilovoltage (Fig. 54).

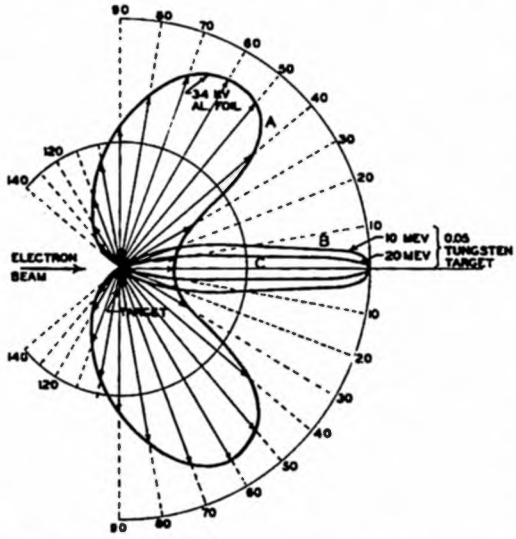


Fig. 54 Polar distribution of X-rays. Reproduced from Johns, H.E., "The Physics of Radiology," 2nd Ed. 1961, Courtesy of Charles C. Thomas, Publisher, Springfield, Illinois.

Simply stated, as electron acceleration is increased, the angle of x-ray production tends to become more forward. At conventional low voltage x-ray energy the x-ray production is approximately at a right angle to the target itself. As the energy increases this angle tends to become more forward. The tube targets are sloped in these machines to take advantage of this difference in direction of x-ray production.

The transmission type target is relatively thin so there is only moderate filtration of the primary beam as it passes through the target. The target (anode) is usually grounded in these tubes. It can then conveniently be made hollow and be either oil or water cooled. The targets characteristically are made of either gold or tungsten in these tubes.

C. Resonant Transformer Units

The first supervoltage x-ray machines, developed in 1939, were of this general design. These machines were originally built to operate in the range between 1 and 2 million volts. In recent years a number of companies have brought

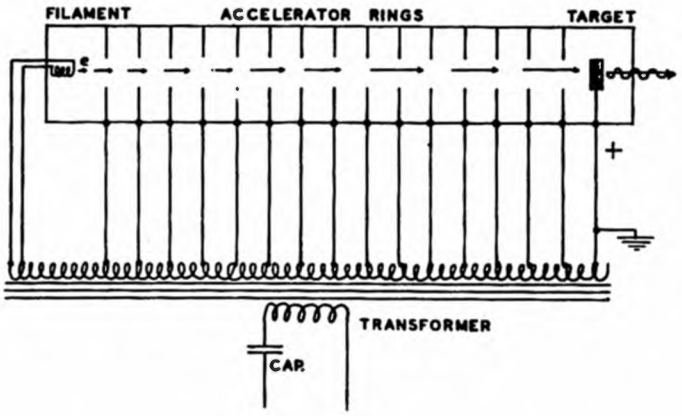


Fig. 55 Resonant Transformer Unit

out machines in the 300 kilovolt range which also utilizes the resonant transformer principle. These machines are small and light weight, when compared to the bulkier conventional orthovoltage treatment machines. The resonant transformer machine utilizes exactly the principle which its name implies, that is resonance.

Resonant Circuit:

A resonant circuit in its simplest form consists of a coil or inductance connected across a capacitor (Fig. 56-a) If a charged capacitor is connected across an inductance a current will be set flowing from the capacitor into the inductance which builds a magnetic field around the inductance. When the energy of the capacitor has been expended creating the magnetic field there is no longer a sustaining current flow and the field collapses. As the magnetic field collapses it generates a current flow in the inductance which then recharges the capacitor. The recharged capacitor is then again able to reform the magnetic field about the inductance and the cycle continues to repeat itself. If the circuit is truly a resonant circuit, the tank energy from the electric field of the capacitor will be shuttled to the magnetic field of the inductance, back and forth. This ocillatory motion is eventually dampened or slowed on each discharge cycle because of resistive losses within the circuit. A current set flowing in this circuit will continue to oscillate until it eventually dies out. The frequency of oscillation in a circuit of this nature is determined by the values of the capacitor and the inductance. A circuit of this nature will have a natural resonant frequency at which it will tend to oscillate.

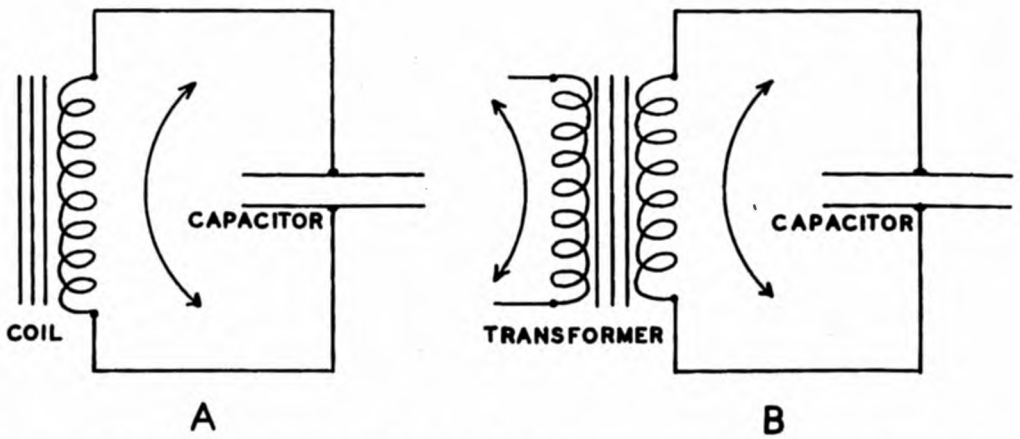


Fig. 56 A. Resonant Circuit. B. Inductance coupled resonant circuit.

If the Inductance is now made into a transformer by the addition of a primary winding(Fig. 56-B),power can now be added by inductance coupling into the circuit to compensate for power lost as the charge slowly dies out in the circuit. If the driving voltage coming in on the primary winding is of exactly the same frequency as the natural frequency of resonant oscillation of the capacitor and inductance, it will require only a very small boost of power from the primary during each oscillatory cycle to keep the current flow in the secondary circuit at a maximum value. Such a circuit is then known as a resonant frequency driven resonant circuit.

The major advantage of such a circuit is that it is now possible to build up a very high voltage in the secondary winding with minimal power being required to keep the circuit at a high potential. This is ideally suited to x-ray work. Out of this design came the resonant transformer type of x-ray unit. (Fig. 55) Initially these units were driven with 180 cycles per second alternating current. The original transformers were very large and quite bulky, due to the heavy windings and heavy iron cores that were necessary. The lighter modern units utilize 1200 cycles alternating currents to drive the resonant circuit. This results in a much smaller and more compact unit for the same equivalent amount of power generated. The secondary of this transformer is tapped at increments to the accelerator rings of the x-ray tube as shown in Fig. 55. Electrons generated at the filament are accelerated down the complete length of the tube in ever increasing increments as they pass the accelerator rings. These rings are so designed that they not only accelerate the electrons, but keep them focussed into a tight beam, eventually striking the target at full energy. A grounded transmission type target is used in these machines.

D. Van de Graaff Electrostatic Accelerator.

The Van de Graaff accelerator is an electrostatic high voltage generator, developed originally in 1933. Van de Graaff generators utilize the principle that a charge deposited upon a smooth spherical surface will concentrate on the exterior of the surface. Van de Graaff generators were originally utilized experimentally for radiation therapy in 1940, following the basic designs of R.J. Van de Graaff. Machines have been developed since that time with capabilities and operating characteristics from 2 MeV to as high as 10 MeV.

In this machine a negative charge is sprayed onto a fast moving non-conductive motor driven belt. This charge is carried very rapidly to the top of the tower where sharp pointed collecting electrodes actually brush the charge off the belt and conduct it to the sphere which caps the top of the tower. Because of the nature of charge collection on spherical surfaces, the charge very rapidly and readily transfers to the outer surface of the sphere, enabling more and more charge to be swept off the fast moving belt. Connected to the sphere is a high resistance column, eventually ending at the base of the tower in a ground connection. These resistances are connected in series with direct connections across to the x-ray accelerator tube which lies parallel to the resistance column. The accelerator rings of the x-ray tube are connected to the junctions between the series string of dropping resistors. The entire unit is so constructed that the potential falls uniformly from an extremely high negative potential at the top of the tower to a zero or ground potential at the base.

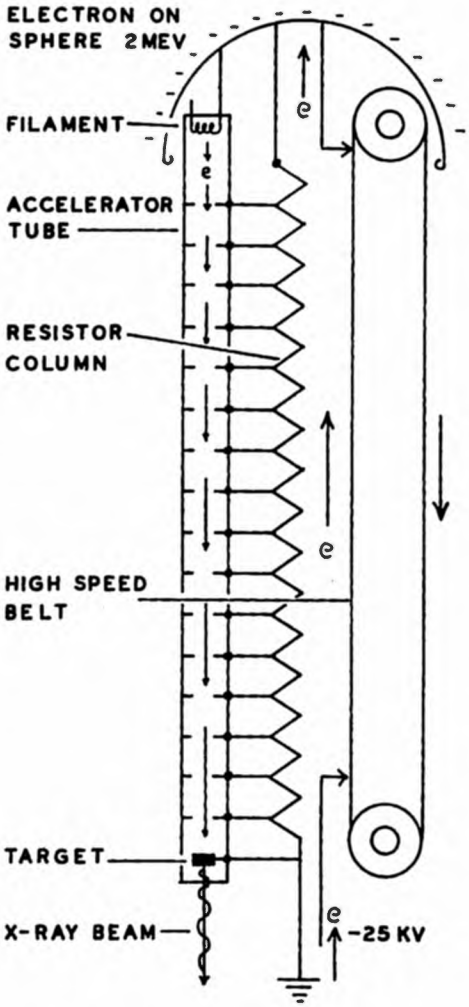


Fig. 57 Van de Graaff Generator

The basic principle of x-ray production is exactly the same as it is with other accelerator tube type units. Electrons are produced by the hot filament at the top of the x-ray tube and are gradually accelerated by the ever decreasing negative potential (the same effect as an ever increasing positive potential) from the top of the tower to the final connection at the grounded anode. The target is a transmission type target with the majority of x-rays being generated in a forward direction. These machines utilize a gold, water cooled, transmission type target.

Van de Graaff generators, as well as the transformer type units, are generally housed in a large pressure tank which is filled with many atmospheres of pressure of dry inert gas, usually a mixture of carbon dioxide and nitrogen. This allows the high voltage to be contained with much less likelihood of corona or arcing.

E. Linear Accelerator

Linear accelerators have arisen as a direct outgrowth of high frequency radar developmental design, which gave the capabilities for the development and control of ultra-high frequency (about 100 megacycles per second) power. These machines utilize the resonant frequency principles previously mentioned. At these high frequencies transformer type circuits are no longer applicable and the actual waves themselves are bound within radiofrequency resonant cavities and are forced to oscillate. Linear accelerators work on the principle of electrostatic attraction and repulsion by alternating positive and negative charges on acceleration plates within the wave guide tube. (Fig. 58)

The principle is quite similar to that used in the accelerator tubes previously mentioned. In the linear accelerator resonant traveling waves of radiofrequency power are applied to the interior of the wave guide tube. These traveling radiowaves passing up and down the tube alternately charge positive and negatively, plates which are unequally spaced within the tube (Fig. 58), with the spacing becoming wider as the electron beam approaches the target end of the tube. The radiofrequency wave which is traveling through this tube creates an alternating positive and negative electrostatic charge on the plates. If electrons are injected into the tube at the proper timing of the cycle they will

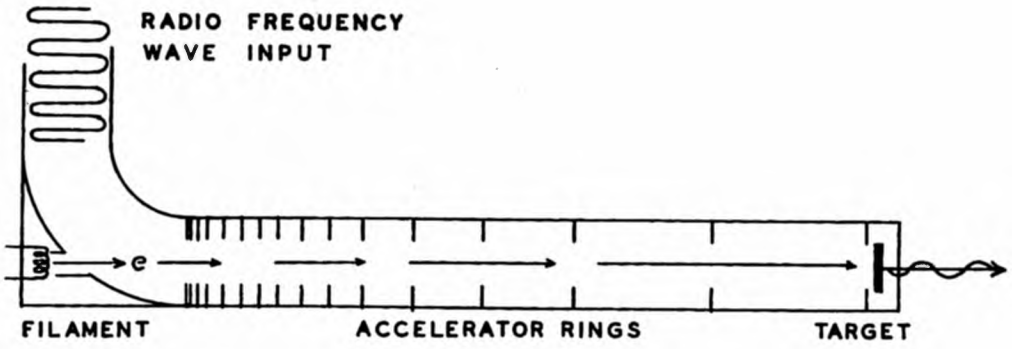


Fig. 58 Linear Accelerator. Adapted from Glasser, O., "Physical Foundations of Radiology," 3rd Ed. 1961, Courtesy of Paul Hoeber, Publisher, New York, N.Y.

be attracted to the first set of plates while they are positive,once the electron reaches the area of the positive electrode and immediately as it passes through it, the charge reverses so that this electrode is now negative and the next electrode down the tube is positively charged. The electron is then attracted toward the next electrode which immediately switches charge just as the electron passes it. Each time this maneuver occurs, the electron gains momentum and energy. Because the driving frequency which is charging the plates is of a constant frequency, and the electrons are continuously moving faster as they pass each successive group of plates the spacing on the plates becomes wider as the target end of the tube is reached. This process of charge reversal occurs throughout the full length of the tube until the electron is traveling at its maximum speed, at which time it strikes the target at the end of the tube.

As in the Van de Graaff generator, the electron beam may be used to generate an x-ray beam by bombardment of a gold target. The electron beam itself may also be ejected from the tube through a special window and be useful as a beam of high energy electrons.

The usual linear accelerator used in Radiation Therapy operates through the voltage range of about 4 to 8 MeV, although experimental units have been built to as high as 50 MeV for medical purposes and to as great as 1000 MeV for physics research.

To this point we have described machines in which the kilovoltage rating was determined by the kilovolts actually developed within the machine. Electrostatic machines and transformer type units, of necessity, must have the actual physical voltage developed within the machine to accelerate the electrons. Linear accelerators do not require that the voltage within the tube be developed to the point of the effective operating x-ray output. As mentioned before energies up to as high as 50 MeV for medical purposes have been developed, however, these are only apparent voltages. The electrons are accelerated through the tube and eventually gain energy of such an amount as to be equivalent to an electron accelerated at that physical electrostatic voltage, although this voltage never existed within this particular tube.

The greatest advantage of the linear accelerator over some other high energy x-ray generators is their high output. The average 4 MeV linear accelerator may produce as high as 300 roentgens per minute at 1 meter treatment distance. It should be noted however, that the dosimetry at this particular energy is very similar to Cobalt 60.

F. Betatron

The first betatron was built by Professor D.W. Kerst at the University of Illinois in 1940. Considerable work was done in 1948 by Dr. G.D. Adams and others in adapting the betatron for medical radiation therapy. The original betatron had an energy of 2.3 MeV. The average medical betatron in use today operates at 22 MeV, with experimental medical units up to 40 MeV and physics research units with energies as high as 300 MeV. As in the linear accelerator, the specifications in millions of volts is the sum of voltage gradients through which the electrons being accelerated are continuously falling; the millions of volts are not an insulation problem as they are never actually developed in their total physical dimensions.

In conventional high voltage transformers the magnetic fields developed by the primary windings are used to accelerate the electrons trapped within the secondary windings. These machines depend upon multiple windings in the secondary to produce the necessary high voltage. In the betatron the secondary winding is replaced by an evacuated tube called the donut (Fig. 59). The electrons are injected into the donut from a hot filament and suspended within a powerful magnetic field developed between the pole faces of a large electromagnet. They are forced to circulate and are accelerated as the magnetic field increases.

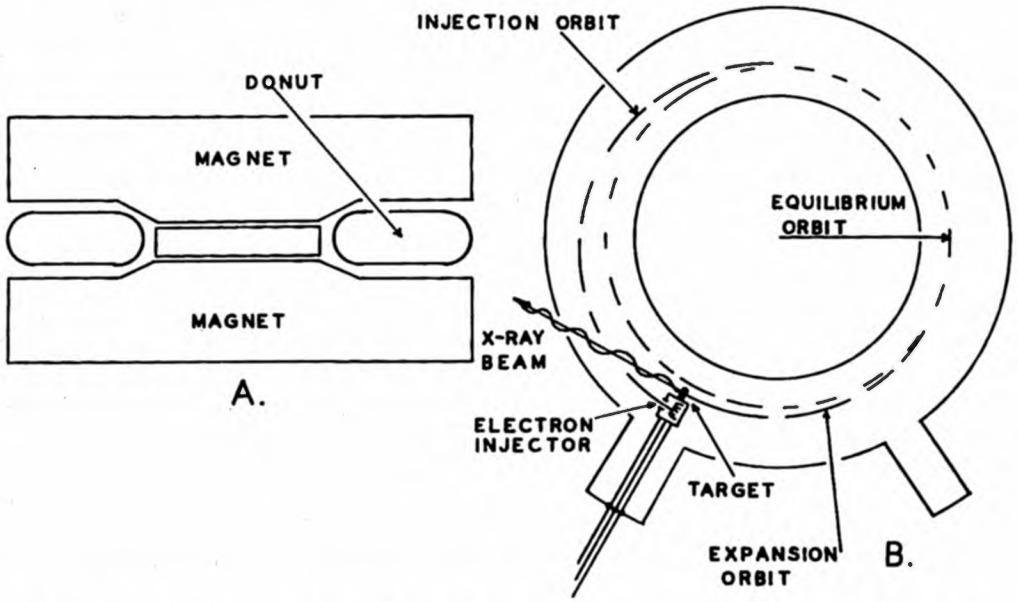


Fig. 59 Betatron. A. Sagittal section through the pole pieces and donut accelerator tube. Fig. B. Top view showing the details of a donut Adapted from Johns, H.E., "The Physics of Radiology," 2nd Ed. 1961, Courtesy of Charles C. Thomas Publisher, Springfield, Illinois

The magnetic field is driven by an alternating current of 180 cycles per second. The electrons are forced into a circular path and spiral slowly inward until they reach an equilibrium orbit which balances magnetic deflection force against centripetal force. The electrons are continuously accelerated as they circulate in this powerful magnetic field due to the rising magnetic flux. In this manner enormous energies are acquired by the electrons during the time of the first quarter cycle of the alternating current.

If the electrons were allowed to remain in the tube through the second quarter cycle the decreasing magnetic field would decelerate them back to rest. As the magnetic field reversed direction on the third quarter cycle, the electrons would be forced to move in the opposite direction, hence a second beam would be generated in opposition to the first. This technique is used on some betatrons.

The machine is designed so that when the field reaches its maximum value a pulse of current is passed through a special coil very near the radius in which the electrons are traveling. This warps the magnetic field which causes the electrons to move out of their equilibrium orbit and strike the target, producing the beam of x-ray that is emitted from the tube. Under special circumstances the circulating electrons maybe led to a thin aluminum window rather then the target and thereby are caused to appear outside the betatron as a concentrated stream of electrons.

Because of the nature of alternating current fields, the betatron is used throughout only the first quarter cycle and remains idle the next 3 quarters of a cycle until the sine wave is returned back to the original correct polarity. At this point another group of electrons is injected, accelerated, expanded, and allowed to strike the target to produce more x-rays. The radiation from a betatron is produced in pulses spaced 1/180 of a second apart. These pulses last about 1 microsecond.

The efficiency of x-ray production rises markedly as the electron voltage increases. The average x-ray efficiency in a diagnostic x-ray machine is on the order of .2% x-ray with 99.8% of the electron energy being wasted in heat. In multimillion volt machines, such as the betatron, the x-ray production may exceed 50% efficiency. No specific cooling is needed for the target, which is usually a small thin sheet of either gold, platinium or tungsten within the donut of the tube.

The enormous speed which the electrons eventually reach in this machine can be appreciated by the fact that throughout one quarter of one cycle or 1/720 of a second, the electrons are accelerated and spiraled through the 15 inch diameter donut approximately 350,000 times, traveling a total distance of about 250 miles.

G. The Synchrotron

The synchrotron was developed to overcome some energy limitations of the betatron. In the betatron the electrons are accelerated to nearly the speed of light by magnetic induction. To create high electron energy by betatron acceleration principles requires a marked increase in the physical size of the magnet which is an absolute necessity for the betatron magnetic fields. By adding a second feature to the betatron, namely, a resonant cavity within the donut, a synchrotron is created. (Fig. 60).

Each time the electrons pass through the resonant cavity, they will be given an increase in energy by means of the same electrostatic principles as described in the linear accelerator. Unlike the linear accelerator where the electrons are forced to be accelerated down a linear tube, which has certain distance limitations, the synchrotron overcomes this by utilizing a circular betatron type of orbit.

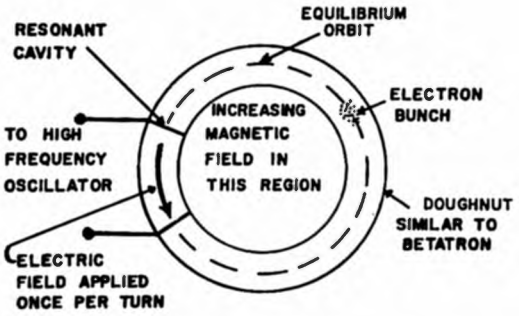


Fig. 60 Synchrotron, schematically represented. Reproduced from Johns, "The Physics of Radiology," 2nd Ed. 1961, Courtesy of Charles C. Thomas, Publisher, Springfield, Illinois.

With this arrangement the electrons are held in a tightly circular field and are forced to pass through the resonant cavity for many revolutions. During each pass through the resonant cavity the electrons are given an added burst of energy.

A Medical Electron synchrotron is available operating at the 70 MeV level. Research synchrotrons have been built with energies much higher than this. Proton synchrotrons have been built with energies up to 28 BeV.

A given size electromagnet is capable of accelerating electrons to a much greater energy when adapted as a synchrotron rather then when operating as a betatron. Whereas, betatrons can be designed to accelerate electrons to at least 300 MeV, the weight saving features of the synchrotron in achieving a desired electron energy has made its evaluation and adaptation to the medical field a practical necessity.

There is still some mystery about the operation of the very high frequency utilized in the resonant cavity. Clearly, the electrons to be accelerated require a minimum potential difference from which energy can be extracted on each transit.

The 70 MeV synchrotron operates at 60 cycles per second. During each cycle the electrons are accelerated to about 1.2 MeV by betatron action. At the proper time the high frequency oscillator power is switched on and the electrons are accelerated in successive transits through the resonant cavity as the magnetic fields confining the orbit also increases, the two being exactly in step.

The electron beam is forced to intersect with a platinum target and a powerful beam of x-ray is produced. A variable energy electron beam may also be produced from the synchrotron.

H. Isotope Teletherapy units

1. Radium

The original teletherapy machines utilized enormous (at that time) sources of radium as their source of radiation. A machine was assembled in the past with

a collection of 10 grams of radium in a single source. This was worth, at that time, approximately twenty million dollars. This machine was assembled as an experimental unit and was not used for an appreciable length of time. Because of the disadvantages of inverse square law and the still relatively small activity and output, these machines utilized a treatment distance of 10 to 12 cm. Due to tremendous expense and other problems incurred by the use of radium, the entire teletherapy project was abandoned until 1951.

2. Cobalt 60

In 1951 two 1,000 curie sources of Cobalt 60 were made available and two Cobalt 60 teletherapy units were constructed to utilize these sources. Cobalt 60 is an artificial radioisotope produced from Cobalt 59. Cobalt 59 is bombarded by neutrons in a heavy water atomic pile reactor. The neutron bombardment of Cobalt 59 creates Cobalt 60 which then undergoes beta decay with the concurrent production of the two gamma rays of 1.17 MeV and 1.33 MeV (See section I-B-1) This gives a nearly monochromatic radiation with a half-value layer of 12 millimeters of lead, and compares very favorably with radiation produced by 3 MeV x-ray machines. The teletherapy machines currently being manufactured at the present time utilize sources ranging from small (1,000 curie) sources to large (10,000 curie) sources. These machines use a variety of source skin treatment distances (SSD). Some of the units utilizing short SSD (24 cm.) are used for low depth-dose, head-and-neck and palliative therapy. Other units, using an SSD up to at least 100 cm, are used for deep therapy, with the deliverance of high tumor doses.

3. Cesium 137.

Cesium 137 is a fission by-product of the operation of atomic piles. Cesium 137 is produced as a breakdown product of uranium decay and as such is termed a reactor "poison", which, along with other breakdown products interferes with the efficiency of operation of the pile. Periodically the fuel elements must be cleaned and the Cesium 137, as well as other breakdown by-products must be removed. These products can then be refined and the various isotopes separated out. Cesium 137 is acquired in this fashion in relatively large amounts. After removal and purification, the cesium 137 may be fashioned into various types of sources. It is not only useful for teletherapy sources but also as a substitute for radium in interstitial uses (III-E). Cesium 137 has a lower effective energy than Cobalt 60 (662 MeV monochromatic gamma emission). Cesium 137 has a half-life of 30 years, this contrasted to Cobalt 60 with a half-life of 5.26 years. (See Fig. 16) The lower energy of Cesium 137 makes it of value for utilization in short treatment distance machines with a 10 to 15 cm SSD. Cesium 137 may in the future replace the conventional ortho-voltage (250 KVP) x-ray machines. Its lower energy is a distinct disadvantage in large SSD units where the greater depth dose of Cobalt 60 is superior. The half-value layer of Cesium 137 is approximately 5.5 mm of lead, which simplifies shielding.

Teletherapy units are extremely simple in their theory and operations. These machines are much simpler and far less involved than the super voltage therapy machines which we have previously discussed.

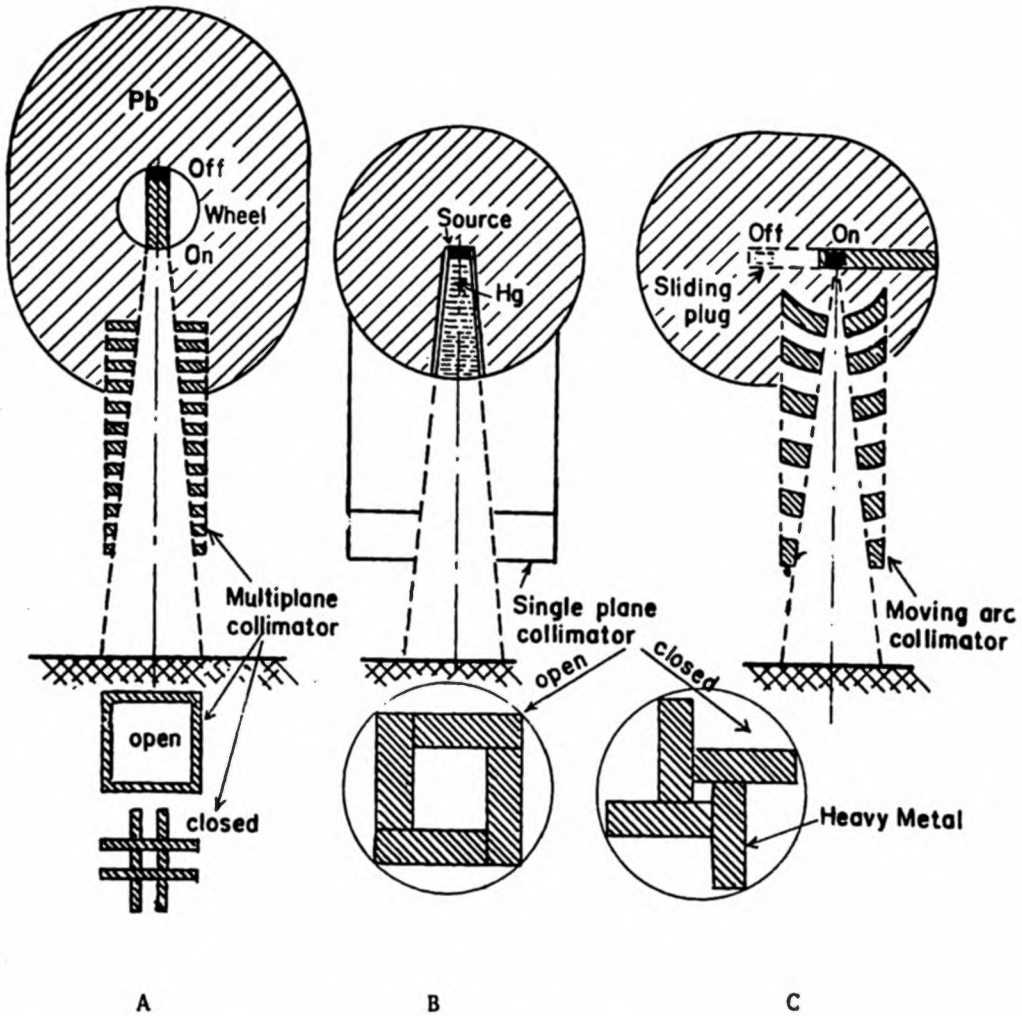


Fig. 61 Mechanism of Teletherapy Units. Including Collimator design, Reproduced from Johns, H.E. "The Physics of Radiology", 2nd Ed. 1961 Courtesy of Charles C. Thomas, Publisher, Springfield, Illinois.

Teletherapy units, unlike conventional x-ray machines, have their source of radiation producing continously. The radiation is continously being emitted from the source, although in ever decreasing amounts in accordance with its half-life (II-A). Because of the continuous "on" condition of the source of radiation, some-method must be devised to confine the radiation within a safe shielding until such time as it is needed.

A multitude of designs were tried on the early units, however, these are now narrowed down to two basic designs with a third design still found on many of the older units. (See Fig. 61) In illustration (A) note that the source is rigidly contained on one side of a movable rotating wheel. With the source in the "off" position it is confined within the central portion of the head and is completely shielded from the exit port by the high-density tungsten-alloy wheel. In the "on" position the source rotates around until it is exposed in front of the opening of the machine.

A method which was used earlier as an alternative to the rotating source wheel utilized a heavy liquid metal (mercury) which was alternately pumped in and out of a chamber (Illustration B). The machine was effectively turned on and the radiation was allowed to escape from the source when the chamber was empty. As can well be imagined there are many problems with this particular arrangement, and although this is found on a number of the older machines, it is not currently being used as a shutter mechanism.

A third method in use is the sliding source drawer or piston. (Illustration C). In this type of machine the source is mechanically arranged in a sliding drawer. In the "off" position it is safely stored within the protective portion of the head. When the machine is turned on the source is pushed, usually by the mechanical action of compressed air to the "on" position where it is adjacent to the opening in the machine head, and its radiation is then used for treatment.

Teletherapy machines in most cases, are absolutely silent during their "on" operating position. Because of this absolutely silent operation, these machines must be designed to "fail safe". In the event of power failure, safety interlock failure, or other operation which would accidentally expose personnel to the "on beam", these machines must automatically return to the source "off" position. All of the currently manufactured units have mechanical indications as to source position.

ACKNOWLEDGEMENT

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